DEVELOPMENT AND APPLICATION OF A PROTOCOL FOR FINE-WIRE AND SURFACE EMG DATA COLLECTION AS PART OF CLINICAL GAIT ASSESSMENT

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Declaration

I declare that this PhD thesis has been composed by myself and embodies the results of my own course of study and research whilst studying at The University of Salford from October 2012 to June 2016. All sources and material have been acknowledged.

Abbreviations

CGA	Clinical gait analysis
CMC	Coefficient of multiple correlation
CV	Coefficient of variance
EMG	Electromyography
ICC	Intraclass correlation coefficient
SEM	Standard error of measurement
VR	Variance ratio

Conference proceeding

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In: 1st Clinical Movement Analysis World Conference, 29th September – 4th October, Rome, Italy [Oral presentation]

Abstract

Background: Electromyography (EMG) is a measure of neural activation to muscles and as such can give a window into neuromuscular dysfunction in patients. Although it was the primary focus of early clinical gait analysis (CGA), it has become progressively less common since the widespread adoption of optoelectronic measuring systems capturing three dimensional kinematics and kinetics. This is surprising since EMG has considerable potential to explain gait deviations observed in the kinematic and kinetic data.

Apart from the extra time required for collecting data there are a number of barriers to the use of EMG in modern CGA. The most obvious is that EMG data has traditionally been collected, analysed and, most importantly, presented using quite different techniques which prevents a streamlined integration of EMG with the kinematic and kinetic data. Secondly, although the general characteristics of normative EMG patterns in the larger muscles are reasonably well understood, there is considerably less consensus on those which are smaller but still clinically important. Finally several of the most clinically important muscles, such as the tibialis posterior (TP), can only be accessed using fine-wire techniques and there is no consensus on how such data should be presented clinically.

Objectives: This research aims to define rigorous data capture, analysis and presentation protocols for incorporation of both fine-wire and surface EMG measurements into CGA. The secondary aim is to provide definitive normative EMG profiles in the selected lower limb muscles across the gait cycle in healthy adults as reference for CGA purposes. Finally, a case series aim to explore whether the methods of collection, analysis and data presentation established in this work could be used to detect patterns of muscle dysfunction underlying kinematic impairments in the gait of stroke participants.

Methods/results/discussion: A systematic review was conducted and the synthesised EMG profiles with and without between-subject variability from all included papers showed a

wide range of variability in lower limb EMG profiles, a lack of studies in deep muscles which potentially play important roles in gait such as TP, no standardisation of fine-wire EMG acquisition and processing (compared to the surface EMG) and various methods of EMG normalisation. These variety of collection and analysis techniques resulted in large variability, in the current literature base, of EMG profiles between different studies. The majority of EMG studies currently available in the literature focus on larger superficial muscles. Studies on TP were scarce in spite of its important role in foot posture and gait. One reason for the lack of information on deep lower limb muscles may be that these can only be assessed using fine-wire sensors, for which there are no guidelines for standardised collection procedures amenable for use in CGA.

A series of experiments aimed at addressing these limitations of fine-wire EMG in the current literature base (identified in the systematic review) and ultimately using improved collection and analyses techniques to allow direct comparison between fine-wire and surface EMG and provide a normative database for clinical application were carried out on TP for which little normative reference data exists, tibialis anterior (TA), and medial gastrocnemius (MG). The normalisation study mean normalisation appears to be the best method to reduce variability and this is true across muscles, sensors and different measures of variability: standard deviation can be reduced by 18%-62% of the mean signal and standard errors of measurement can be reduced up to 42% of the mean signal. A peak normalisation is equally effective with small difference (<5%). The second study revealed six gait cycles were necessary to collect fine-wire EMG which showed similar patterns (r >0.9) at the same standard error of an ensemble average of surface EMG for TA and MG. The grand ensemble average of fine-wire EMG showed slightly greater between-session variability than surface EMG (9%-10% for fine-wire and 4%-7% for surface). Normative EMG data was then

collected using normalisation with respect to the peak over six gait cycles from TP, TA and MG alongside kinematics and kinetics at five different speeds from eight young participants. Finally a case series of EMG collections with participants with stroke were used to explore the proof-of-concept of how standardised EMG methods could be implemented in clinical gait analysis and the potential benefits of using EMG to support identification of reasons for gait deviations in CGA. A normative database collected using these established methods was effective to identify pathological features and changes of muscle activity in three participants with post-stroke when using ankle-foot orthosis (AFO). However, the sensitivity of the database to detect changes under AFO condition depended on the severity of the impairment.

Conclusion: As there was no previous standardised guidelines for the use of fine-wire EMG in CGA, this PhD defined a protocol for EMG measurement of TP, TA and MG using fine-wire and surface sensors in combination with kinetics and kinematics for CGA. The results of a series of systematic examinations of different normalisation techniques as well as between subject and between-session variability indicate that six gait cycles of data is sufficient for the collection of fine-wire EMG in CGA and that normalisation relative to the mean or peak during the gait cycle is the most appropriate if EMG data is to be used to aide CGA. A case series of stroke participants demonstrated data collected in this way could be used to detect impaired muscle activation underlying impaired kinematics of walking when compared to a normative database, and that the EMG data could add useful information to understanding typical CGA outputs.

Chapter 1 Overview and scope of the thesis

1.1 Electromyography (EMG) in clinical gait analysis

1.1.1 Role of EMG in the management of neuromuscular disorders

a) Importance of neuromuscular disorders

Neuromuscular disorder is a broad term describing a range of conditions that affect the function of muscles as a direct or indirect consequence of pathology affecting the central or peripheral nervous system. Most of these disorders result in the individual affected having difficulties in moving around which vary in age of onset, severity, and speed of progression. Common neuromuscular disorders include stroke, cerebral palsy (CP) and Parkinson's diseases. The estimated prevalence rate of the 24 most common neuromuscular disorders was 160 per 100,000 population from literature reported in 2015 (Deenen, Horlings, Verschuuren, Verbeek, & Van Engelen, 2015). In the UK, there were over 70,000 diagnosed cases from a population of over 60,000,000 (Pohlschmidt & Meadowcroft, 2010). Overall in the UK, the stroke incidence is about 152,000 a year (Townsend et al., 2012). The major causes are ischaemic which accounted for the majority (85%) and haemorrhagic (15%) (Feigin et al., 2013).

b) Importance of clinical gait analysis in managing neuromuscular disorders

Patients with neuromuscular disorders frequently present with complex walking patterns due to primary and secondary deficits and the selection of appropriate treatment can be difficult. Scientific approaches utilising objective and quantitative measurement equipment can be effective to identify problems in the assessment and treatment monitoring processes of patients with neuromuscular disorders (Kleissen, Buurke, Harlaar, & Zilvold, 1998). The use of gait analysis for children with CP is the most common and has been in use in specialist centres for over 30 years. Kinematic, kinetic and EMG data are analysed preoperatively to

inform clinical decision making (particularly choice of procedures during complex orthopaedic surgery) and postoperatively to quantify its effects (Gage, Perry, Hicks, Koop, & Wernt, 1987; Gueth, Abbink, & Reuken, 1985; Hoffer, Barakat, & Koffman, 1985). The careful use of gait analysis has led to improvements in surgical techniques (Gage, 1993) and other treatments such as ankle-foot orthosis (Tyson & Kent, 2013).

c) Importance of EMG with clinical gait analysis

Electromyography (EMG) can be regarded as an essential tool in gait analysis, which primarily focusses on lower limb extremities for patients with motor disorders (Kleissen et al., 1998). It is regarded as the only direct method to identify the pattern of muscle dysfunction in an individual (Perry, 1992). The technique can be used for monitoring in clinical follow-up and evaluation of the treatment.

1.1.2 History of EMG within clinical gait analysis

a) Early studies of gait analysis and muscle function

Gait analysis has been a topic of interest since the time of Aristotle (Baker, 2007). The first major development occurred during the renaissance in Europe as mathematical expressions were used for the first time to describe movement by Rene Descartes (Baker, 2007).

Leonardo da Vinci and Andreas Vesalius showed interest in the analysis of muscles and their functions as seen in their anatomical drawings. Their work was, however, limited to the dissection of dead muscles as noted in Basmajian and De Luca (1985). Muscle-generated electricity was first observed by Italian Francesco Redi in 1666 (Basmajian & De Luca, 1985). Giovanni Borelli (1685) was the first researcher to investigate the forces generated by muscles and acting through tendons. Modern electrophysiological studies started in 1792 when Luigi Galvani (1792) illustrated that the muscle contraction could be triggered by electricity as cited by Kleissen et al. (1998).

Jules Etienne Marey is regarded as the first modern gait analyst to apply Newton's laws to the human body (Baker, 2007). With his student, Gaston Carlet, he measured the vertical component of the ground reaction using pressure transducers in 1872 which led, with another student, to the development of a pneumatic force plate (Marey, 1883). Marey also experimented with chronophotography using a range of different types of markers to obtain exquisite images for clinical interpretation (Baker, 2007). Marey was also the first researcher to record the electrical activity during a voluntary muscle contraction and introduced the term 'electromyography'in 1890 (Clarys, 1994).

Willhelm Braune and Otto Fischer conducted three-dimensional gait analysis using four cameras and a continuous exposure, with the participant walking in the dark with flashing Geissler tubes strapped to the body (Braune & Fischer, 1987). The forces acting on each segment were calculated. This is the foundational work for modern gait analysis.

b) Post-war pioneer of clinical gait analysis

After the second world war, a group established at Berkeley carried out comprehensive biomechanical gait analysis to facilitate design of prostheses by using bone pins to accurately measure the movements in transverse plane and minimize skin movement artefacts (Eberhart et al., 1947). Perhaps surprisingly, given their use of bone pins, they considered that EMG was too invasive to use. They emphasized that the improvement in mobility treatments including surgery, physiotherapy, and prostheses relied heavily on the understanding of normal locomotion.

Between 1960s and 1970s EMG played an important role in clinical gait analysis, probably, because it was easier to measure than three-dimensional data. Inman et al. combined electromyography (EMG) recorded from indwelling sensors with three dimensional force and energy measurement to gain insight into normal gait (Inman, Ralston, Todd, &

Lieberman, 1981). However, the available technology was limited and resulted in labour intensive and time consuming data processing.

Jacqueline Perry and David Sutherland added other gait data to aid interpretation. Perry introduced both observational and instrumented methods for measuring temporal-spatial data (Perry, 1992). While Sutherland continued developing the three dimensional data on five joint angles using three cameras (Sutherland & Hagy, 1972). Later Jurg Bauman synchronized EMG data with the images taken from movie files (Baumann & Baumgartner, 1974). As a result of advanced camera technology, clear images of the foot taken from a transparent foot plate resulted in useful reference data on foot movement during gait.

c) Rise of 3-D kinematics and kinetics

With the arrival of computerized data processing, the time taken to process data was dramatically reduced and routine assessment of multiple data stream became possible. Sensors with direct interface to the computer were developed to monitor movement of body segments, exemple of such as sensors including the electrogoniometer and accelerometers (Karpovich, Herden, & Asa, 1959; Lamoureux, 1971; Morris, 1973).

In 1967 Furnée presented a study of arm movement using a video camera based opto-electronic system (Furnée, 1967) and a little later Dinn, Winter, and Trenholm (1970) independently developed a similar video based technology. Such systems were used extensively by Winter, Greenlaw, and Hobson (1972), and Jarrett, Andrews, and Paul (1976) and laid the foundation for current technology which allows synchronisation and calculations of kinematics, kinetics, EMG, and energy consumption in real time.

1.1.3 Current status of EMG within clinical gait analysis

Many gait analysis laboratories are currently equipped with EMG. However, there is considerable variability of EMG methodology and reporting across the different

laboratories. EMG data may be presented in the form of raw, rectified or linear envelope signals or as being on or off with respect to some arbitrary threshold. By contrast there is a much more standardised approach to the presentation and reporting of kinematic and kinetic data. This difference may reflect the relative importance currently attributed to data from different sources. Kinematic and kinetic data are now generally regarded as the core measurements and thus standardized presentation and reporting techniques have evolved. EMG, by contrast, is regarded as of secondary importance and been subject to less standardisation. This has established a vicious circle as, without standardized methods of reporting, it becomes more difficult for a consensus to develop as to how EMG data should be interpreted which may further lead to a degradation in its perceived importance relative to other data sources.

A key aspect of interpretation of data during clinical gait analysis is the comparison of data from multiple sources. Consistent formatting of kinematic and kinetic data allows ready comparison of what is happening at different joints, in different planes and between joint angles, moments, powers and components of the ground reaction. Standardised methods of presenting data from a number of trials and summary data (either as a representative or average trace) allow a consideration of the variability in movement patterns alongside that of the representative trace. Recent suggestion that gait graphs should be marked-up with symbols (Baker, 2013) also makes biomechanical interpretation much more transparent but is only practical if presentation of gait graphs is in a similar format to that currently regarded as standard for kinematics and kinetics. There is thus a strong argument for developing methods of capturing, processing and presenting EMG data in a format that is directly comparable to the current standards for kinematic and kinetic data.

For clinical purposes, definitive normative EMG profiles are required in order to diagnose individual patients (Winter & Yack, 1987). There are numerous studies describing

normative data but they have employed a range of techniques for data acquisition and processing including normalisation techniques and analysis often with inadequate justification for selection (Agostini et al., 2010; Kadaba et al., 1989; Lyons, Perry, Gronley, Barnes, & Antonelli, 1983; Schwartz, Rozumalski, & Trost, 2008; Sutherland, 1984, 2001; Wootten, Kadaba, & Cochran, 1990; Yang & Winter, 1985). In particular, EMG normalisation is a critical technique to allow comparison between subjects and sessions. Several techniques have been proposed but there is no consensus on which should be used clinically (Burden, 2010). Differences between acquisition and processing techniques affect the EMG profiles making the use of available reported data as clinical reference complicated. This also causes difficulty in clinical interpretation and comparisons between different laboratories. And whilst it is standard practice for laboratories to collate their own normative reference databases for kinematics and kinetics this is much less common for EMG data.

Though several guidelines on EMG have been published (Hermens & Merletti, 1999; Merletti, 1999), there are many substantial issues to be resolved before the standardised procedures and normative data can be established. For example, many patients who are diagnosed with rheumatoid arthritis or neurological disorders walk with impaired speeds. This suggests that provision of the normal EMG profiles over a range of speeds will be beneficial (Den Otter, Geurts, Mulder, & Duysens, 2004). Furthermore, the majority of research is done on superficial muscles because they can be accessed non-invasively with surface electrodes. Deeper muscles such as the tibialis posterior (TP) are often clinically important as well but there is very little data published describing normative patterns of EMG.

This PhD research therefore aims to address these issues first by developing a standard method of capturing and processing fine-wire and surface EMG data and, secondly,

presenting the data in a format compatible with the kinetic and kinematic data used in clinical gait analysis. A secondary aim is to provide a definitive normative EMG dataset in selected lower limb muscles. This along with three dimensional kinematic and kinetic data across the gait cycle in healthy adults as reference for CGA. EMG of the deep muscle tibialis posterior is also focused upon as it may be impaired in patients with neuromuscular disorders and contribute to gait deviations to aid treatment plan.

The key research questions are:

Chapter 3

i) Is there a consensus on profiles of lower limb muscle EMG during healthy walking and on protocols for capturing and processing EMG signals during gait?
   a. Does the level of between-subject variability across studies decrease according to the quality of reporting/quality?
   b. Is there a consensus on EMG profiles of lower limb muscle during gait and what is the between-subject variability of them?
   c. What are required from further investigations?

Chapter 5

ii) In healthy participants, what effect does the normalisation method have on the between-subjects and between-sessions repeatability of linear envelope EMG signals collected with fine-wire or surface sensors?

iii) In light of this and other differences between the normalisation schemes, which is the most appropriate for future clinical use?

Chapter 6

iv) How sensitive are fine-wire EMG signals to where the sensors are placed?

v) How many more gait cycles of fine-wire EMG are required to give the same confidence that the mean is representative compared with surface EMG?

vi) How do EMG signals from surface and fine-wire sensors compare between sessions (repeatability) and across a range of different walking speeds?

Chapter 7

vii) How EMG data from both surface and fine-wire can be integrated more effectively within the conventional methods of clinical gait analysis in healthy adults.

viii) How does walking speed affect the activity of the muscles of the lower leg and associated kinematics and kinetics?

ix) Is there any evidence that the neuromuscular changes associated with ageing are manifested in the EMG of the lower leg muscles during walking?

Chapter 8

x) Does the adoption of such methods for data collection, processing and presentation of EMG data enhance the clinical interpretation of gait analysis data?

xi) Is the variability around normative reference EMG collected using our suggested protocols sufficiently low to detect differences in the EMG profiles of the tibialis posterior, tibialis anterior and medial gastrocnemius due to neuromuscular pathology (stroke)?

xii) What does the EMG report add to an analysis of kinematics and kinetics in CGA?

-Can EMG profiles help to explain the gait deviations seen in kinematic and kinetic graphs between normative data and participants with post-stroke?

-Can EMG profiles show the difference between walking with and without AFO for participants with post-stroke?

1.2 Rationale of the thesis

1.2.1 Chapter 3 Systematic review: Normative EMG profiles of lower limb muscles during gait.

This chapter shows the grand EMG profiles of the lower limb muscles synthesised from a number of studies and determined the available EMG collecting, processing and analysis techniques. The systematic review was conducted using focused search terms on several databases to ensure capture of all relevant data. Customised data extraction and quality evaluation tools based on available relevant guidelines of EMG and research methods (e.g. Hermens and Merletti (1999); Merletti (1999); Von Elm et al. (2008)) were used to ensure the best possible evidence was retrieved. In this review, the importance of EMG measurement guidelines is emphasized in order to obtain consistent EMG data. This review identified a number of issues in EMG measurements: lack of reported activity in deep muscles (despite the importance in gait), several normalisation techniques used and technical issues of fine-wire EMG. These formed the basis for further investigations in this PhD work.

1.2.2 Chapter 5 Normalisation of EMG during normal gait

This is the first study to analyse the between-subject variability and the between-session variability of fine-wire and surface linear envelope EMG of the same muscles (tibialis anterior and medial gastrocnemius), and fine-wire EMG of tibialis posterior, normalised using different techniques. Normalisation of EMG data is different from other normalisation schemes as it is to reduce variability in the magnitude of the signals detected by sensors that arise from a variety of factors: the positions of sensors, muscles, individuals and

sessions/days (Burden, 2010) in order to elicit variability in the underlying muscle responses. Common techniques were compared: peak, mean and maximal voluntary isometric contraction (using a dynamometer), which in theory provides clinical meaningful data. This led to a recommendation of the most appropriate normalisation technique to be used in the standardised protocol, then applied in subsequent chapters.

1.2.3 Chapter 6 Application of fine-wire sensors for EMG measurement of tibialis posterior in clinical gait analysis

This chapter addresses some of the challenges regarding the application of fine-wire sensors in the measurement of the EMG of the tibialis posterior. This has important clinical implications since the muscle potentially makes an important contribution to gait deviations in participants with neuromuscular deficits, yet little is known about the activation of this muscle during gait. The similarity and variability (between sites of the same muscles, between subjects and between sessions and when walking at different speeds) between fine-wire and surface EMG were compared to determine protocols of collecting and processing of the fine-wire EMG in CGA. This is the first study to investigate the sensitivity of fine-wire in tibialis posterior and the comparisons between two types of sensors on tibialis anterior and medial gastrocnemius.

1.2.4 Chapter 7 Incorporating EMG data with kinematic and kinetic data to understand walking in healthy adults

This study used protocols for capturing and processing both surface and fine-wire EMG to aid understanding of healthy adult gait: effects of speed and effect of age along with kinematics and kinetics. These two issues are found in CGA as patients with neuromuscular disorders may walk at wide range of self-selected speeds: slower or faster than self-selected speed from healthy adults. In addition, many patients such as stroke are older than those

recruited in normative reference. In order to identify the changes due to purely to different walking speeds or older age from those due to pathology, the dataset consisting of EMG, kinematic and kinetic data were presented with SD across the gait cycle and compared for the first time. The results will serve as a guide to provide a local normative database for CGA.

1.2.5 Chapter 8 Demonstration of EMG measurements of the tibialis posterior in clinical gait analysis

This case series demonstrates the protocol for collecting and processing fine-wire and surface EMG with other gait data using an impairment focused approach. The comparison is of gait datasets between (1) stroke patients and healthy adults walking at matched-speed and (2) stroke patients walking with and without an AFO. The comparisons will explore if EMG data can help explain kinematic and kinetic gait deviations and provide proof-of-concept for the use of EMG in standard CGA practice.

1.2.6 Source of data for different chapters

Three samples of participants were recruited for the different studies and different sub-sets of these used to answer different research questions. The association between each sample and studies are illustrated in Figure 1.2.

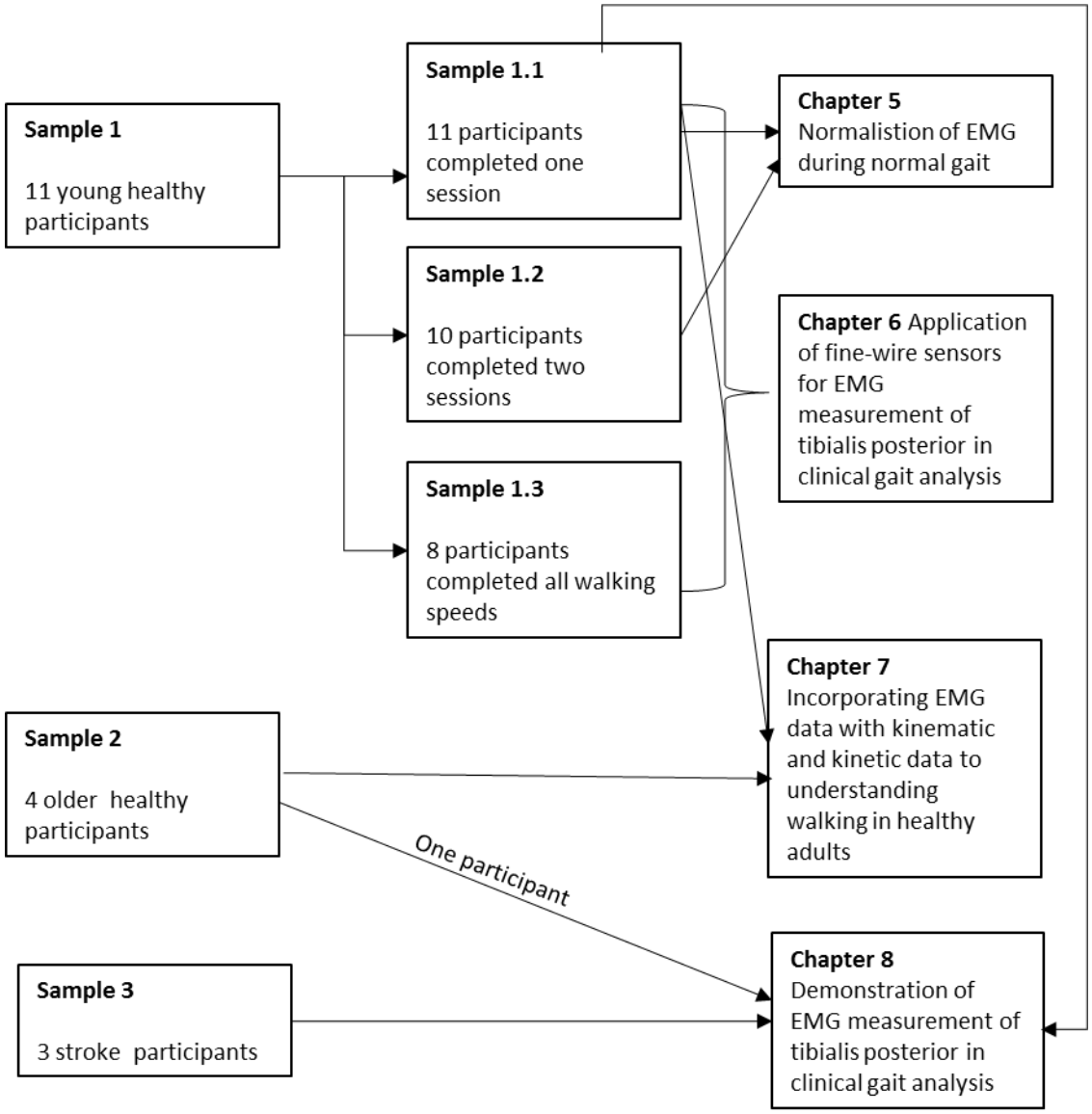


Figure 1.1 Sample recruited for overall thesis

1.3 Potential significance of this research

This PhD research will fill the gaps in the EMG gait literature which included (1) a lack of normative EMG from deep muscles (despite their importance in gait) and (2) needs for standardized protocols to aid CGA which are applicable to both fine-wire and surface sensors. Through the series of experiments, standardised protocols will be developed. There will be a specific consideration of the different properties of fine-wire and surface EMG which might require different processing methods in order for data to be considered comparable.

A key achievement of this work will be the development of protocols for fine-wire EMG collection during dynamic task-gait. Whilst developed specifically for tibailis posterior the techniques will be applicable to all deep muscles which potentially play important roles in gait such as flexor hullucis longus, extensor digitorum longus and to superficial muscles which are subjected to crosstalk from neighboring such as rectus femoris, adductor magnus and peroneus longus.

The overall significance will be to make available to researchers and clinicians set of standardised data collection, processing and presentation methodologies which allow both fine-wire and surface EMG to be reported in a manner consistent with existing conventions for the presentation of kinematic and kinetic data in clinical gait analysis. This will also have application to other muscle types involved in neuromuscular disorders. The clinical utility of this new approach will be illustrated by a number of case studies also carried out within the scope of this research.

Chapter 2 Background

Electromyography is the recording of electrical activity in the muscle regarded as 'a valuable technique for studying human movement, evaluating mechanisms involving neuromuscular physiology, and diagnosing neuromuscular disorders'(Kamen & Gabriel, 2010; Robertson, Caldwel, Hamill, Kamen, & Whittlesey, 2013). This involves recording, processing, and displaying electromyographic signals. *Electromyographic signal/electromyogram (EMG)* is the name given to the total signal detected by a sensor. 'The EMG signal is the algebraic summation of motor unit action potential trains from all active motor units within the pick-up area of the electrodes' (International Society of Electrophysiological Kinesiology. Ad Hoc Committee, 1980). The EMG signal is generated wherever a muscle is activated, regardless of whether this is voluntary or involuntary (Robertson et al., 2013).

EMG is attractive in movement studies because it provides a window into physiological processes which cause a muscle to generate force and produce movement and behaviour. At this time, it remains one of our only direct windows into the neural codes that produce muscular contraction, force, and movement. EMG has been used for over 20 years now as an evaluation tool for medical research, rehabilitation, ergonomics and sport science.

2.1 Motor unit action potential

2.1.1Generation of muscle fibre action potential

Motor unit is the term used to describe the single smallest controllable muscular unit. The motor unit consists of a single alpha-motor neuron, its neuromuscular junction and the muscle fibres (as few as 3, as many as 2000) (Basmajian & De Luca, 1985; Enoka, 2008; Gath & Stålberg, 1981)(Figure 2.1*). An alpha motor neuron* has its cell body in the anterior horn of the spinal cord and innervates a group of muscles fibres. Once activated by the central nervous system or reflex, the motor neuron transmits *the action potential (MUAP)*

along its axons. When this reaches a synaptic end bulb, it triggers a sequence of electrochemical events to release the neurotransmitter –acetylcholine (Ach). This chemical crosses the synaptic cleft, binding to the receptors situated in the motor end plate of the muscle which consists of Ach receptors. The activation of the Ach receptors causes an influx of sodium ions and efflux of potassium ions resulting in a depolarisation of the postsynaptic membrane and leading to an action potential.

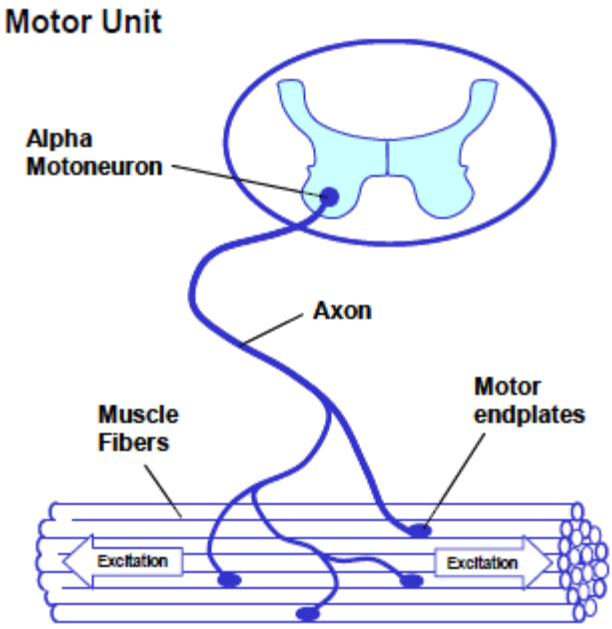


Figure 2.1 Motor unit (Konrad, 2006)

During resting, the ionic equilibrium is maintained by an active ion pump and forms a resting potential of around -80 to -90 mV (Tortora & Derrickson, 2008). The resting potential can be affected by exercise training and types of muscle fibres: fast or slow twitch (Hammelsbeck & Rathmayer, 1989; Kamen & Gabriel, 2010; Moss, Raven, Knochel, Jr, & Blachley, 1983). The changes in membrane potential: depolarisation, overshoot, repolarisation, and after-hyperpolarisation, are described as action potential (AP). There may be an after-hyperpolarisation phase, during which the membrane potential is temporarily more negative than the resting level (Figure 2.2).

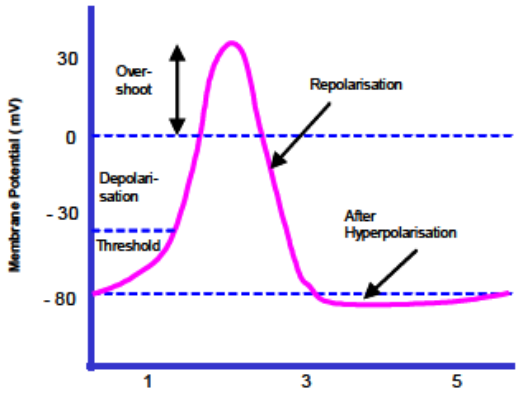


Figure 2.2 The action potential (Konrad, 2006)

Two types of the voltage gated channels play important roles (Figure 2.3). The voltage-gated sodium ion channels embedded in the sarcolemma open, allowing sodium ion influx and causing depolarization if a certain threshold level, around -30 mV, is exceeded. During depolarisation, the negative membrane potential becomes less negative, reaches zero and becomes positive, up to +30 mV. Then the potassium ion channels open up, allowing the potassium ions to flow out. This reduces the positive membrane potential-depolarizing phase. When the potassium ion channels remain open after the repolarising phase ends, the after-hyperpolarisation occurs (Tortora & Derrickson, 2008). Then the membrane is refractory when sodium ion channels are returning to an active state and the potassium ion channels are closing. These take a few milliseconds but it is unlikely that the other action potential can occur. The generated action potential propagates along the excitable membrane of the muscle fibre, triggering the release of the calcium ions into the sarcoplasm and the muscle fibres subsequently contract as a result of electro-mechanical coupling. Regarding this relation, it is assumed that, in a healthy muscle, any form of muscle contraction is accompanied by these aforementioned mechanisms (Tortora & Derrickson, 2008).

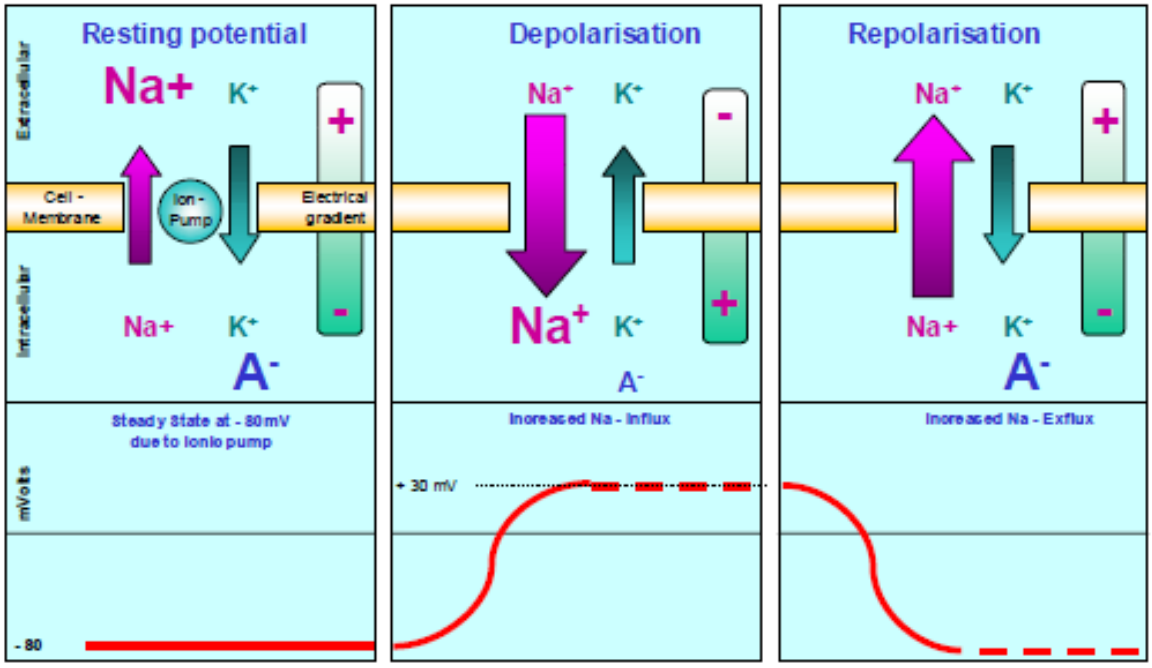


Figure 2.3 Depolarisation/repolarisation within excitable membrane (Konrad, 2006)

2.1.2 Propagation

The action potential arises at the motor end plate and, as well as triggering electromechanical couplings, propagates towards both sides of the fibres along the sarcoplasmic reticulum and transverse tubule for the deeper portion of the fibres. This allows the simultaneous activation and contraction of every part of the muscle fibres (Tortora & Derrickson, 2008). The longer axon of the motor neurons of certain muscle fibers may cause a slightly delayed activation in those fibers (Robertson et al., 2013).

2.1.3 Motor unit action potential

The number of muscle fibres innervated by a motor neuron is termed the *innervation ratio* (Kamen, 2004). *The motor unit action potential (MUAP*) is the summation of electrical activity of all muscle fibres activated within the motor unit. It is proportional to the innervation ratio. The smaller motor units are usually recruited before the larger ones until the required muscular force has been achieved –*Henneman size principle* (Henneman, Somjen, & Carpenter, 1965). Also, the increase in the number of recruited motor units

increases the strength of the contraction. Furthermore, the frequency of the activation, firing rate or motor unit discharge is controlled by the nervous system. The higher frequency produces a higher muscular force (Robertson et al., 2013).

Motor unit activation initiates muscular force. A force time response to an action potential is a twitch. It is relatively common for a motor unit to receive a number of action potentials, resulting in overlapping twitches (Enoka, 2008). The summation of these twitches at peak level results in a force plateau profile - *a tetanus*. The tetanus is unfused or incomplete when the activation rate is insufficiently high, resulting in oscillating forces around the average level. Moreover, doublets and synchronisation are the processes that can alter the muscular force. The *doublets* are two short bursts of motor unit activation before firing at a regular rate to generate a greater force than the usual two twitches (Clamann & Schelhorn, 1988). This strategy is used when considerable effort is required to initiate limb movement. *Synchronisation* is a process whereby more than one motor unit fires simultaneously.

The EMG signal is the electrical summation of all of the active motor units in the detecting volume. The signal consists of both negative and positive components. The amplitude of the recorded EMG may reflect the intensity of the muscular contraction to some extent but the relationship between the amplitude of the recorded EMG and the muscular force is frequently non-linear (Solomonow, Baratta, Shoji, & D'ambrosia, 1990).

2.2 EMG acquisition

The recording arrangement can be either *monopolar* or *bipol*ar. In a monopolar arrangement, the sensors are on the muscle belly and the electrical reference point on the bony prominence. This method is frequently used in isometric contraction (Ohashi, 1995, 1997)and investigations using needle electrodes (Dumitru, King, & Nandedkar, 1997). It is probably more susceptible to movement artefacts (Robertson et al., 2013). In a bipolar

arrangement (single differential), two electrodes are placed on the muscles and a third neutral is grounded. The electrical difference between the two electrodes is amplified. The common signal from these electrodes is probably attenuated-*common mode rejection*. The recommended minimum common mode rejection ratio is 100 decibels (Robertson, 2004). The gained EMG output should be in the range of the analogue-digital converter.

2.2.1 Sensors

There are two main types of electrodes: surface or skin electrodes, and inserted (wire and needles) electrodes (Basmajian & De Luca, 1985). Surface sensors are widely used because they are non-invasive, cause minimal discomfort, and are reasonably reproducible (Jacobson, Gabel, & Brand, 1995a)(Figure 2.4). Surface sensors record the EMG signals generated by a number of individual motor units within a large volume under the detection surface and therefore are prone to artefacts caused by the movement of the muscle's innervation zone towards the detection volume during a dynamic contraction (Bogey, Perry, Bontrager, & Gronley, 2000; Rainoldi, Melchiorri, & Caruso, 2004).

a. b.

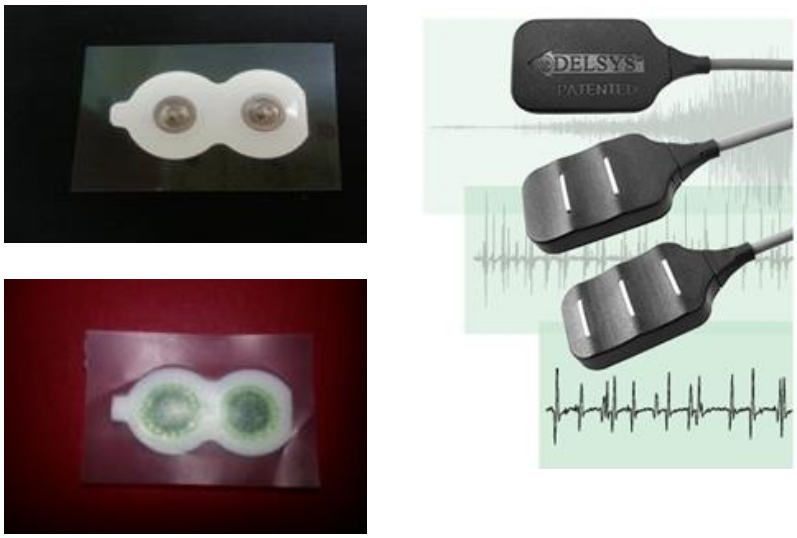


Figure 2.4 Surface sensors a) Passive sensor, b) active sensors (Delsys Inc., 2015)

The surface sensor consists of the metal conducting surface connecting to the amplifier. The surface sensors can detect EMG signals generated from a number of individual motor units within the pick-up area of the detection surface. They are available in active and passive forms (Figure 1.4). The passive form consists of a detection surface which detects the current on the skin through its skin-electrical interface. The active form increases the input impedance, so it less sensitive to the impedance of the electrode and skin interface.

The passive form consists of a detection surface which detects the current on the skin through its skin-electrical interface. The removal of hair and the dead surface layer of the skin through light skin abrasion can lower the electric impedance (Tam & Webster, 1977).This can be done by using abrasive gel and cleaning with an alcohol wipe. Continuous pressure on sensors over the skin provided by adhesive strips or collars is also recommended (Basmajian & De Luca, 1985).

The application of a saline gel or paste on the electrode-skin interface also improves the detection of the electrical signal. At the metal-electrolyte junction, the chemical equilibrium may be altered by a number of factors; for example, changes in temperature, changes in electrolytes concentration of the paste or gel, relative movement of metal and the skin, and the flowing current. As a result, the polarization potential is altered. The utilisation of chloride with the metal sensor provides a reversible chloride exchange. This arrangement can diminish the polarisation potential which is associated with the sensors (Basmajian & De Luca, 1985).

The active surface sensor increases the input impedance, so it is less sensitive to the impedance of the sensor and skin interface. The active sensor does not require skin preparation or an electrical medium. This type of sensor has an operational amplifier close to the detection site. This arrangement results in a ‘cleaner’ signal because of the high signal to noise ratio (SNR) (Hagemann, Luhede, & Luczak, 1985). They are classified into two

types: resistively coupled or capacitively coupled to the skin (Basmajian & De Luca, 1985). The latter is not applicable to EMG application because of its poor reliability. When resistance is achieved in the order of 10^{12} ohm, an adequately large input impedance is achieved. The application of an active sensor is relatively convenient.

The disadvantages of both types of surface sensors include the susceptibility to crosstalk from adjacent muscles, and the inability to detect specific deep muscles. The surface sensor can effectively record the signal within a 10-20 mm distance from the surface layer (Barkhaus & Nandedkar, 1994; Fuglevand, Winter, Patla, & Stashuk, 1992).

Furthermore, the surface electrodes can be designed as an array depending on the specific research purpose, such as a longitudinal array. The array can consist of a grid of nine or more electrode surfaces to investigate the architectural features of the muscles (Thusneyapan & Zahalak, 1989).

Inserted/indwelling sensors: a fine-wire and needle sensors involve invasive procedures, so they are not frequently used in gait analysis to detect EMG patterns during walking for the superficial muscles. A fine-wire sensor consists of the two small insulated wires with bared tips (Figure 2.5). They are available in a single wire or as two wires. These wires are threaded through a hypodermic needle for insertion with the tips bent back to form a barb to retain the sensor in the muscle when the needle is withdrawn. The distance between the bared tips determines the detecting volume. The other end of the wires connects to the amplifier.

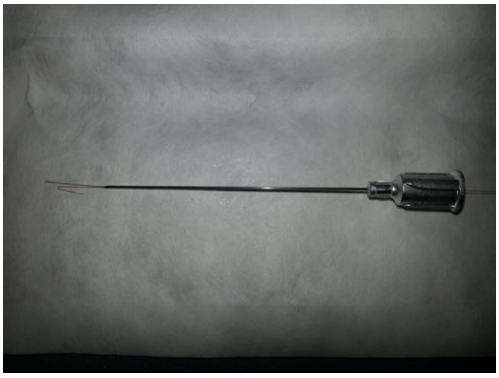


Figure 2.5 Paired-hook wire sensor

Fine-wire sensors are implanted in the muscle of interest and only record from a small volume of muscle and therefore suffer less from crosstalk. This technique also enables the investigation of deep tissue (Bogey et al., 2000) and may be minimally affected by the movement of the muscle beneath the skin during a dynamic activity such as gait. However the signal from a small volume may not represent the activity of the entire muscle (Bogey et al., 2000).

Needle sensors of various designs incorporate unipolar or bipolar sensors into the hypodermic needle which remains in place for measurements. Modern manufacturing allows for very small inter sensor distances and hence very specific signal detection. They can be painful when walking and are highly sensitive to movement. The use of the needle sensor in gait analysis has not been found in this body of work.

2.2.2 Electrode placement

A number of guidelines have been published: Anatomical Guide For The Electromyographer: the Limbs And Trunk (Perotto & Delagi, 2005) describing the locations for needle sensor, the SENIAM project: Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles (Hermens & Merletti, 1999), Recommendations for the standardisation of lead positions in surface electromyography (Zipp, 1982), and Output forms Data analysis and application (Leveau & Anderson, 1992). Many recent papers e.g.

(Barr, Miller, & Chapin, 2010; Den Otter et al., 2004; Hof, Elzinga, Grimmius, & Halbertsma, 2002; Huber, Nueesch, Goepfert, Cattin, & Von Tscharner, 2011; Murley, Buldt, Trump, & Wickham, 2009; Nene, Byrne, & Hermens, 2004; Prosser, Stanley, Norman, Park, & Damiano, 2011) follow the SENIAM guidelines.

There are several principles in determination of sensor locations for recording AP from the muscles. Firstly, the tendinous area and the motor points should be avoided as a tendon is an insulator. Secondly, the motor points are subject to variable EMG signals. When sensors cover the neuromuscular junctions, the recorded signals are the algebraic subtraction from a differential amplifier and may be cancelled. This is because APs are travelling towards both sides of the motor endplate. Therefore, SENIAM projects recommend using the mid position between the most distal motor endplate zone and the distal tendon in the longitudinal aspect of the muscle. Thirdly, the sensors should be parallel to the muscle fibres to avoid the reduction of the signal amplitude and altered frequency. This reduction could be up to 50% (Vigreux, Cnockaert, & Pertuzon, 1979). Fourthly, the sensor should be positioned away from the edge with other subdivisions or muscles to avoid the presence of a signal from the adjacent muscles. Finally, the distance between the sensor pairs or inter-electrode distance is also important and should be kept constant between sessions. Vigrex et al. (1979) found that an inter-electrode distance of 60 mm can detect the greatest amplitude of EMG signals when using a 7mm diameter sensor. However, the latter study found no significant changes in EMG amplitude after using different surface sensor types, different recording areas, and different inter-electrode distances (Jonas, Bischoff, & Conrad, 1999). A higher spectral frequency is observed when the sensors are placed closer together (Bilodeau, Arsenault, Gravel, & Bourbonnais, 1990; Moritani & Muro, 1987).

2.2.3 EMG signal processing and analysis

The standard guideline for reporting EMG data suggests that a frequency between 5Hz-500Hz contains most of the surface EMG power spectrum (Hermens & Merletti, 1999), so the data acquisition should cover the signals within that range.

The amplitude and frequency of the EMG signals are the common characteristics of interest. Muscle activation (force generation) is a result of a number of active motor units and the frequency of activations. A higher activation will thus result in an EMG signal with a higher amplitude (number of motor units) and more frequency components (frequency of activation). Conventional EMG analysis, through the signal processing described below, combines these two factors in determining the magnitude of the linear envelope.

a) Amplitude characteristics

There are five major variables describing the amplitude of the EMG signal in this context. Firstly, peak to peak amplitude is useful when the signal is highly synchronous - containing multiple simultaneously firing motor units (Robertson et al., 2013). Secondly, the average rectified amplitude is the average of the absolute alternating current as the EMG signal appears as the interference pattern with a zero average. Thirdly, the root mean square amplitude calculates the square values. Thus, rectification is unnecessary. Fourthly, the linear envelope is an estimation of the "volume" of activity. The EMG signal is a full wave rectified before passing through a low pass filter. Cut off frequencies between 3Hz and 50Hz have been suggested (Robertson et al., 2013). Also 10 Hz shows a satisfactory waveform for short duration activity (Robertson et al., 2013). As the high frequencies content was attenuated from the signal, the remaining signal may be inappropriate for onset offset analysis. Finally, integrated electromyography is the total accumulated activity over a period of time.

b) Frequency characteristics

The frequencies in EMG signal can be described by turning points and zero crossing. This method counts the number of peaks per unit of time and the number of times that the signal crosses the zero level. The latter is correlated with the other frequency variables such as spectral analysis (Inbar, Allin, Paiss, & Kranz, 1986). The mean and median frequency or spectral analysis techniques are frequently used. A positive skewness with an approximated mean and median of 120 Hz and 100 Hz, respectively, is frequently found in surface recorded EMG (Robertson et al., 2013). Changes to these characteristics can be used to indicate the changes in the conduction characteristics of the muscle fibres.

Other common techniques include onset-offset analysis. The onset-offset technique determines the start and end points of muscle activation (Sutherland, 1984, 2001). Thus the EMG signal should not be filtered or processed, which may diminish the high frequencies content. Moreover, numerous techniques have been developed to analyse the EMG signal for specific purposes: recurrence quantification analysis (Filligoi & Felici, 1999), neural network classification(Liu, Herzog, & Savelberg, 1999), wavelet analysis (Karlsson, Yu, & Akay, 1999) as examples. However, in this work, only the time varying amplitude will be considered to be consistent with the other gait data such as kinematics and kinetics.

2.2.4 Normalisation

Normalisation is carried out by dividing the EMG signals obtained from a specific task or event by the EMG signals from a reference contraction of the same muscle and presenting this as a proportion or percentage (Burden, 2010). The application of normalisation enables the EMG data recorded from different subjects, muscles, and days to be compared. This technique was suggested to be used before comparison between treatment conditions (Lehman & Mcgill, 1999).

The recent guidelines provide general information for wide applications of EMG, not specific to clinical gait analysis.The Surface electromyography for the Non-Invasive Assessment of Muscles (SENIAM) project resulted in recommendations for EMG collection and amplitude estimations including spectral analysis for surface EMG and a set of test signals. However, these are limited to superficial muscles. The ISEK standards for reporting EMG require specific information about surface, intramuscular, needle sensors and identify the range of filters to be reported but do not recommend the specific sensor placements, or signal processing (Merletti, 1999). As a result different measuring protocols continue to be used in different laboratories. There are also many factors affecting the obtained EMG signals. The technical factors may include the sensor types, sensor positions, crosstalk, noise, and data acquisition. The physiological factors may include tissue characteristics and the location of muscles such as deep muscles. The noise from the testing environment as well as electrical, motion artefacts, and magnetic power from surrounding machines, as well as walking speed can alter the amplitude and frequency of the EMG signals. Moreover, there are numerous techniques used to process and analyse the data. These variations lead to difficulty in the interpretation and comparison of the EMG.

Chapter 3 Systematic review: Normative EMG profiles of lower limb muscles during gait

Electromyography (EMG) is an essential tool in gait analysis for patients with motor disorders to indicate patterns of muscle activation (Kleissen et al., 1998). It is regarded as the direct method to identify the pattern of muscle activity resulting from neural stimulation and generating muscle contraction and force (Perry, 1992). Patients with motor disorders such as cerebral palsy and stroke frequently present with complex walking patterns and understanding the mechanisms of impairment, through the use of EMG to gain a window into neuromuscular control, can make an important contribution to identifying the best treatment.

Several processing and analysis techniques for EMG are proposed to aid interpretation of the EMG signals: rectified signal, linear envelope, on-off analysis, frequency analysis as examples. This contrasts with the reporting of kinematics and kinetics which is much more consistent in the literature. Within most clinical services, kinematic and kinetic data are plotted against normative reference data (generally by displaying a grey bar representing the +/- one standard deviation range about the mean). EMG however is presented in many different ways with many centres working from short sequences of raw data. It would be useful to provide EMG signals as definitive time-series normative EMG profiles with standard deviation (SD), similar to other gait data for clinical gait analysis (CGA). This would allow clinically relevant information to be consistently presented with other gait data facilitating clinical interpretation. For example, the level of the plantarflexion during swing phase in a graph of kinematics could be a result of inactive tibialis anterior or hyperactive plantarflexors: gastrocnemius and soleus. A standard reporting format would allow data from different studies to be more readily compared or pooled for a normative database.

Whilst it was assumed that robust normative reference data for EMG during healthy walking already exists, a preliminary scoping literature search found much less data than had been anticipated within the peer-reviewed literature and a lack of consensus on what constitutes normal EMG activity. Several studies provide a normative profile of EMG of individual lower limb muscles across the gait cycle but most have used small samples limiting confidence in between-subject variability. A smaller number describe between-session variability. Further there is considerable heterogeneity in data collection and analysis techniques across studies often without adequate justification for selection of methodologies e.g. (Kadaba et al., 1989; Lyons et al., 1983; Sutherland, 1984, 2001; Wootten et al., 1990; Yang & Winter, 1985). Whilst in the past it has been assumed that variation between different laboratories is inevitable there has been recent emphasis on standardizing clinical gait analysis provision to facilitate comparisons of data between different centres (Pinzone, Schwartz, Thomason, & Baker, 2014).

Guidelines for capturing surface EMG, published in response to the large variability in data collection/processing methodology in order to allow data exchange, are now well established (Hermens & Merletti, 1999; Merletti, 1999). The Surface electromyography for the Non-Invasive Assessment of Muscles (SENIAM) project resulted in recommendations for EMG collection and amplitude estimations including spectral analysis for surface EMG and a set of test signals. However, these are limited to superficial muscles. The ISEK standards for reporting EMG require specific information about surface, intramuscular, needle sensors and identify the range of filters to be reported but do not recommend the specific sensor placements, or signal processing (Merletti, 1999). As a result different measuring protocols continue to be used in different laboratories. It is not clear how much consistency exists in the results of data from different measurement protocols and whether any particular EMG profiles over the gait cycle for any particular muscles can be considered normative. This is

compounded by variability in how *healthy participants* are defined and the small sample sizes used in many studies.

The purpose of this chapter is thus to systematically review studies in the peer-reviewed literature which have reported normative lower limb EMG profiles during a gait cycle to 1) examine the level of the variability across studies, 2) synthesise EMG profiles from across studies to determine a normative EMG profile for muscles for which consensus exists and 3) identify muscles for which no such consensus exists and where further investigation is required.

3.1 Research questions

i) Does the level of between-subject variability between studies decrease according to the quality of reporting/quality?

ii) Is there a consensus on EMG profiles of lower limb muscle during gait and what is the between-subjects variability of them?

iii) What are required from further investigations?

3.2 Search method

An electronic search of the literature was conducted in July 2015 using the following databases: AMED (from 1985), CINAHL (from 1982), the Cochrane database of systematic reviews (from 1991), ISI Web of Science (from 1990), MEDLINE using Pubmed and Ovid (from 1996), and SPORTDiscus (from 1985) from their inception to July 2015. A search using MESH terms and free text words was conducted using terms related to "electromyography", "EMG", "gait", and "normal subject". The truncation or wildcard symbols were used to retrieve all possible suffix variations of a root word. (Appendix 1)

3.3 Selection criteria

The aim was to identify studies that provide time varying series measurements of EMG from lower limb muscles, across the gait cycle, during straight walking at self-selected speed in groups of healthy individuals. To achieve this, studies were included if they fulfilled the following criteria:

3.3.1 Participants

Studies were included if they were on healthy adult humans. Where a single group of participants was described as healthy adults (mean age was between 18 and 60), the study was included. In order to fulfil the aims of synthesizing a normative database of healthy EMG profiles during walking, studies on athletes or persons with either medical conditions or abnormal walking patterns were excluded.

3.3.2 Study design

According to the aims of the review, the available normative EMG profiles should be synthesized from best possible studies during walking to minimize the influence of the heterogeneity of collection, analysis and reporting method, which might be expected to differ from studies which aimed to detect differences between conditions. Therefore only studies specifically designed to determine normative EMG patterns collected from the lower limb muscles during walking in straight line without perturbation/obstruction over ground or on a treadmill were included. Biomechanical modelling or computational modelling studies reporting modelled or estimated EMG data were excluded.

Studies that reported/investigated activities of lower limb muscles collected from group of healthy people across a gait cycle were included. A single case design, or case series, or studies with group of healthy people as control for pathology or experimental manipulation on reflexes, or nerve conduction, or any other use of EMG were excluded. Studies involving

experimental manipulation of muscle activation or reflexes were also excluded as were those with a primary purpose of investigating nerve conduction or any similar use of EMG.

3.3.3 Outcome measure

Studies which reported time varying series of EMG amplitude, across the gait cycle, from any of the lower limb muscles, of which muscles crossing hip joint are regarded as upper limit of proximal muscles, were included.

3.4 Quality of measurement and study

Standardised quality assessment checklists such as The Cochrane Handbook for Systematic reviews of Interventions (Higgins & Green, 2009) and Critical Appraisal Skills Programme (CASP) checklists (The Critical Appraisal Skills Programme (Casp), 2013) for randomised controlled trial, diagnostic, systematic review, cohort, qualitative, and case control, are too general to be appropriate for the study designs and outcome measures included in this review. A customised quality checklist was thus designed to facilitate a systematic assessment of quality. This was based on International Society of Electrophysiology and Kinesiology (ISEK) standard for reporting EMG data (Merletti, 1999), the Surface ElectroMyography for the Non-Invasive Assessment of Muscles (SENIAM) European recommendation for surface electromyography (Hermens & Merletti, 1999), the STROBE statement (Von Elm et al., 2008) for the reporting of observational studies in epidemiology and the quality assessment tools of previous review in related areas (Perotto & Delagi, 2005; Ridgewell, Dobson, Bach, & Baker, 2010). A 'Clearly stated objective' was regarded as the primary item in many checklists, this item has effectively been used as inclusion criteria in this study and did therefore not appear in the checklist which contained sixteen questions (Table 3.1). Each question is worth one mark which consisted of several points. All included papers were scored based on the available information that appeared in the published articles

and attached supplementary materials. This checklist was combined with a list of key study characteristics (Table 3.2) to form a single customised data extraction tool.

Table 3.1 Reporting quality scores

	Questions	Scores (%)		
		Pre-SENIAM	Post SENIAM	Total
	General reporting quality	(n=10)	(n=14)	(n=24)
1	Were the participants 'characteristics adequately described?	59	75	68
2	Did they define 'healthy/normal' subject?	70	86	79
	EMG reporting quality			
3	Were the sensors clearly described?	86	87	87
4	Was the location of the sensor adequately described?	10	71	46
5	Was the signal test for crosstalk carried out?	30	36	33
6	Was the detection mode and amplification adequately described?	30	53	43
7	Was the filtering of the raw EMG specified?	47	57	53
8	Was method of analogue rectification described?	70	79	75
9	Was computer sampling adequately described?	55	50	52
10	Was EMG processing adequately described?	88	89	89
11	Was the walking speed stated?	50	100	79
12	Was elimination of gait initiation, turning, and gait termination considered?	70	71	71
	Result reporting quality and validity			
13	Were the graph clearly presented?	100	100	100
14	Was the sample sized justified?	0	14	8
15	Were the EMG result supported by the other literature?	100	100	100
16	Were limitations clearly described?	50	93	75

Note: Pre-SENIAM –the papers published before SENIAM and Post-SENIAM –the papers published after SENIAM

Table 3.2 Key study characteristics

Studies	Year	Acquisition guideline	Quality scores	Sample size	Age range	Sensors types		Normalisation	Conditions	Walking speed (m/s)
						Surface	Fine-wire			
Lyons	1983	Not mentioned	79%	11	25-34	-	☼	Maximal voluntary isometric contraction	Floor	1
Yang	1985	Not mentioned	51%	11	18-33	☼	-	None	Floor	1
Ericson	1986	Not mentioned	61%	10	20-32	☼	-	Maximal voluntary isometric contraction	Floor	1
Arsenault	1986	Not mentioned	53%	8	16-33	☼	-	Maximal voluntary isometric contraction	Floor	106 steps/min
Winter	1987	Not mentioned	61%	19	20-35	☼	-	Mean amplitude during gait cycle	Floor	106 steps/min
Ounpuu	1989	Zipp	50%	10	18-33	☼	-	Mean isometric contraction	Floor	SSW
Pierotti	1991	Perotto and Delagi	61%	15	21-29	☼	-	Peak amplitude during a gait cycle	Floor	SSW
Ciccotti	1994	Not mentioned	42%	22	25-32	-	☼	Maximal voluntary isometric contraction	Floor	2
Davis	1995	Not mentioned	66%	9	29-55	-	☼	Maximal voluntary isometric contraction	Floor	SSW
Olree	1995	Not mentioned	49%	10	21-40	☼	-	Peak amplitude during a gait cycle	Floor	SSW
Hof	2002	SENIAM	79%	20	19-25	☼	-	None	Floor	1
Nene	2004	SENIAM	81%	5	22-33	☼	☼	Peak amplitude during a gait cycle at the fastest speed	Floor	1
Clancy	2004	Not mentioned	58%	15	20-40	☼	-	No	Treadmill	1
den otter	2004	SENIAM	70%	9	17-27	☼	-	Peak amplitude during a gait cycle at the fastest speed	Treadmill	1
Warren	2004	Not mentioned	67%	19	20-38	☼	-	None	Treadmill	1
Nymark	2005	Perotto and Delagi	74%	18	23-58	☼	-	Mean amplitude during a gait cycle	Both	1
Chleboun	2007	Not mentioned	76%	9	20-26	☼	-	Peak amplitude during a gait cycle	Treadmill	1
Murley	2009	SENIAM, Leis and Trapani 2000, Chapman et al 2006	78%	15	12-45	☼	☼	Maximal voluntary isometric contraction	Floor	1
Barr	2010	SENIAM	86%	20	18-37	☼	☼	Maximal voluntary isometric contraction	Floor	1
Bovi	2011	Not mentioned	43%	20	22-72	☼	-	Peak RMS value during the gait cycle	Floor	1
Prosser	2011	SENIAM	75%	10	20-29	☼	-	None	Both	1
Semciw	2013	Semciw et al 2012	76%	15	20-25	-	☼	Maximal voluntary isometric contraction	Floor	1
Murley	2014	SENIAM	82%	30	18-30	☼	☼	Maximal voluntary isometric contraction	Floor	1
Semciw	2014	Semciw et al 2012	72%	15	20-25	-	☼	Maximal voluntary isometric contraction	Floor	1
Total						19	9			

Note: SSW-self-selected walking speed

3.5 Identification of included studies and data extraction.

Titles and abstracts were assessed by a single reviewer (PO) and those that were obviously irrelevant excluded. Two reviewers (PO, KH) independently examined all remaining papers and selected those fulfilling the inclusion criteria. Any disagreements were identified and consensus agreed through discussion.

The same two reviewers also extracted information required for the review using the customized data extraction tool. Again, any disagreements were identified and consensus agreed through discussion.

3.6 Data analysis

EMG data in each included study were digitized by using the open source freeware WebPlotDigitizer (Version 2.5, 2012; http://arohatgi.info/WebPlotDigitizer). The digitised data were then interpolated with spline fills to 100 time intervals across the gait cycle using Matlab (Version 2009a, The MathWorks Inc., Natick, Massachusetts). A study published by Hof et al. (2002) contains both graphical data and an appendix containing the raw data values. This digitization process was thus tested by calculating the RMS difference over the gait cycle between data digitized from the graphs and the raw values. The result, 0.85% of peak amplitude, gave confidence in the use of this method of digitisation. Following digitization all time varying amplitudes were scaled as a percentage of their peak value (across the gait cycle). Combined mean with combined SD were calculated as described below.

There were too few studies for formal assessment but visual inspection of gait graphs from several different muscles suggested little difference in EMG data series between different normalization schemes or walking on the treadmill or floor (Appendix 2 illustrates this with data from the tibialis anterior and lateral gastrocnemius). Therefore all graphs obtained from

identified studies were over-plotted on the same graph after normalization to their maximal amplitudes regardless of data acquisition, original normalization, walking on floor and treadmill, and data processing to enable comparison of EMG profiles between studies and quantify on their agreement.

In order to synthesize the normative EMG profiles for lower limb muscles, the muscles were grouped into 3 different categories based on their availability of reported standard deviation and number of included studies to allow examination of the variability. The categories are (1) those reported by multiple studies with between-subject variability, (2) those reported by a single study with between-subject variability and (3) those reported without between-subject variability.

All time varying amplitudes were normalised by their peak value. The grand mean, which was the mean amplitude of all normalised EMG signals, and grand standard deviation (SD) were calculated considering the number of participants included when they are available. The graph of grand mean amplitude with ±2SD grey areas for each sub-group for all studied muscles were plotted, in a similar format to standard kinematic and kinetic graphs in clinical gait assessment.

The grand mean, *M*, and standard deviation, *SD*, are given by

$$M = \frac{\sum_i n_i \, m_i}{N}$$

$$SD = \sqrt{\frac{\sum_i \left((n_i - 1) s_i^2 + n_i (M - m_i)^2 \right)}{N - 1}}$$

where n_i is the number of observations in each sample with mean m_i, standard deviation s_i and

$$N = \sum_i n_i$$

The SD was calculated from s_i from only the included studies which report SD whereas the mean was calculated from all studies. The graphs of grand mean amplitude with ±2SD grey areas were plotted for each muscle with lines of combined means from both sensors, similar to kinematic and kinetic graphs in clinical gait analysis. For each muscle, the averaged quality score and the averaged timing of peak with corresponding SD were calculated from that of all papers which reported data for that muscle. The contribution of each study to the combine mean was weighted according to the study's sample size (also expressed as a percentage).

To examine the variability of EMG profiles, the averaged SD across the gait cycle was calculated. Then in order to determine whether the level of between-subject variability between studies decrease according to the quality of reporting/quality, only studies which had over 50% of quality score were included to synthesized the grand EMG profiles. This should allow identification of the highest quality data which can be used to derive/identify normative EMG profile from a consensus of the literature and identify the muscles which require further investigation (either a paucity of data or poor quality data).

3.7 Results

There were 8,513 studies identified, of which 24 articles, which presented time series magnitudes of EMG from lower leg muscles in healthy adults over a complete gait cycle, were assessed by two reviewers for data extraction and quality assessment (Figure 3.1, Table 3.1, and 3.2).

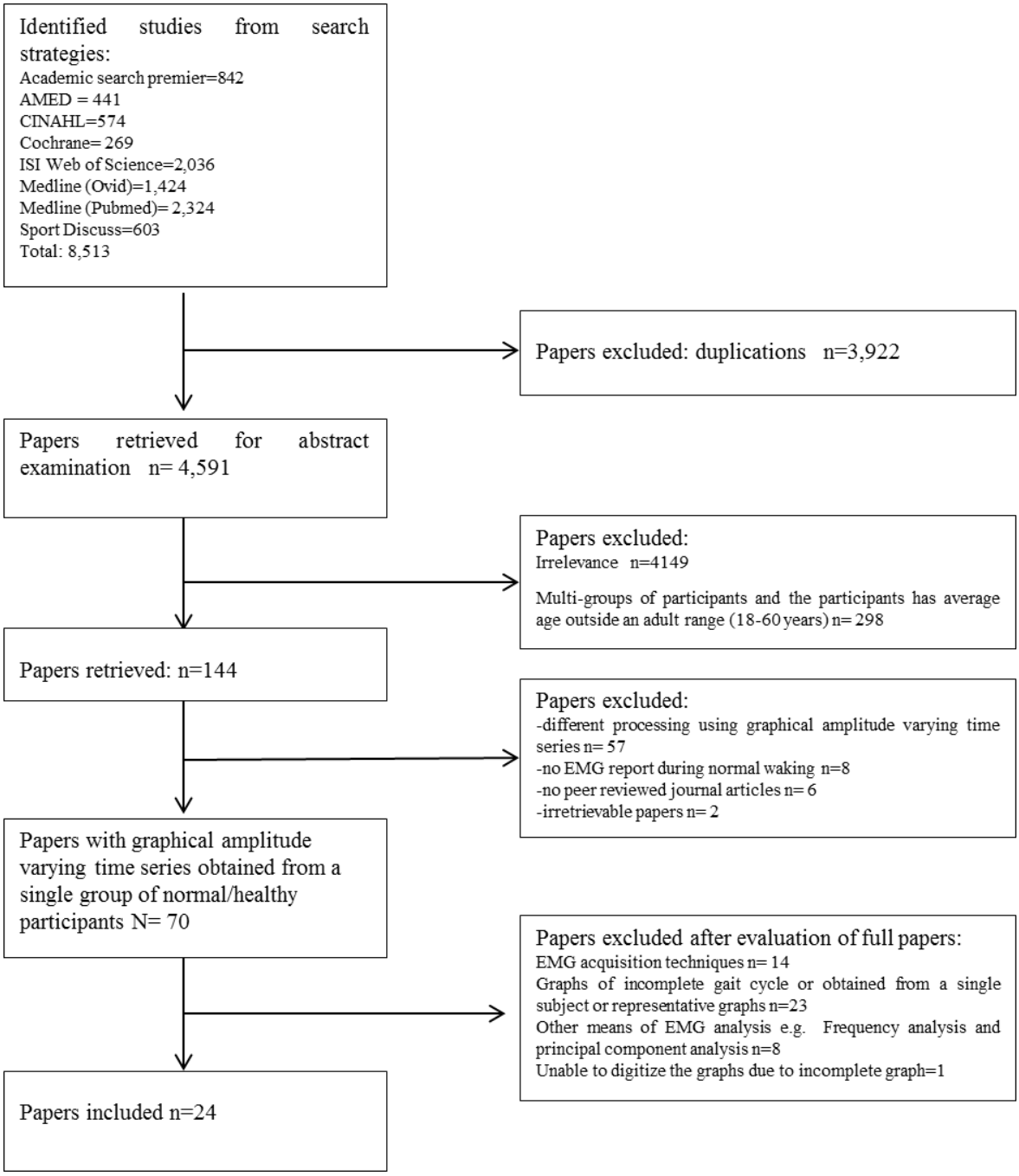


Figure 3.1 Diagram of paper identification

3.7.1 Study characteristics

According to Table 3.1, the average quality score of included papers was 66% (SD 13%). The average was greater for papers published after the introduction of SENIAM guidelines in 1999 (73%, n=14) than it was for those published before (57%, n=10). In particular, the

details on EMG acquisition described in studies published after 1999 was more complete. Studies recruited an average of 14 participants (range 5-30 participants) with an averaged (±SD) age of 28 (±6) years and range of 15-72 years, weight of 70 (±6) kg and height of 176 (±5) cm. Approximately 80% (n=19, involving 273 participants) of identified papers employed surface sensors on the superficial muscles. The muscles most commonly investigated (in over 100 participants and at least 9 studies) are vastus lateralis, rectus femoris, medial gastrocnemius, soleus, tibialis anterior using surface sensors. There were only a relatively small number (n=9, involving 142 participants) of studies that used fine-wire sensors to investigate superficial and deep lower limb muscles. The muscles which were investigated by fine-wire sensors only were tibialis posterior, popliteus, tensor fascia latae, upper and lower compartments of gluteus maximus, anterior and posterior compartments of gluteus medius and glueteus minimus.

Some form of magnitude normalisation was employed in 19 studies in this review (Table 3.2). The denominators included maximal voluntary isometric contraction (MVC) (10 studies), mean amplitude during an isometric contraction (1 study), mean amplitude during a gait cycle (2 studies), peak amplitude (6 studies) during a gait cycle. 4 included studies reported speeds as cadence and "self-selected speed" without giving the actual speed. For those studies reporting speed, the average was 1.1 m/s but ranges from 0.83 m/s-1.50 m/s. Only 5 studies examined walking on a treadmill as opposed to over ground.

3.7.2 EMG profiles and variability

Due to the availability of reported between-subject variability of EMG profiles from different studies, the grand EMG profiles were allocated to one of three groups:

(1) those reported by multiple studies with between-subject variability (n=16)(Table 3.3-4)

(2) those reported by a single study with between-subject variability (n=5) and

(3) those reported without between-subject variability (n= 9) (Table 3.5)

Group 1: Muscles reported by multiple studies with between-subject variability

Table 3.3 and 3.4 describe the variability in the pooled data from across all included studies of the 16 muscles from Group 1 and shows how this is affected if studies with lower quality scores (quality score<50%) are excluded. This decreases the variability (SD) substantially in some clinically important muscles (by between a quarter and a fifth for gluteus medius, lateral hamstrings, soleus and tibialis anterior) with little effect on most muscles and just three (adductor magnus, lateral gastrocnemius and peroneus longus) showing slight increases (1-2%). Excluding the lower quality studies affected the timing of peak activity but generally by a small amount (1-5%). In two muscles (vastus medialis and bicep femoris), the SD of Arsenault et al. (1986) was large because of the method of normalization used, so this study was removed from the grand mean to prevent misinterpretation as between-subject variability.

There were too few studies of fine-wire sensors for most muscles for much to be learnt from comparison with data from surface sensors. Removing data taken from fine-wire sensors had little effect on SD of grand EMG profile for any muscle other than peroneus longus for which it reduced variability by about a fifth. It did however make considerable differences to the timing of peak activity for both peroneus longus and rectus femoris. Similarly there were too few treadmill studies to draw anything conclusive from comparison with over ground walking with no clear pattern of effect on variability between studies emerging when the small number of treadmill studies were removed.

Table 3.3 Analysis table for reported muscle Group 1

Muscles (studies)	With all included studies				Without studies with QS < 50%			
	Studies (N)	Average QS (%)	Timing of Peak (%)	Average SD (%)	Studies (N)	Average QS (%)	Timing of Peak (%)	Average SD (%)
Gluteus maximus [Ericson et al. (1986),Winter et al. (1987),Olree et al. (1995),Hof et al. (2002),Bovi et al. (2011)]	5(65)	56	5(2)	35	3(35)	65	6(1)	28
Gluteus medius [Lyons et al. (1983),Ericson et al. (1986),Winter et al. (1987),Olree et al. (1995), Hof et al. (2002)]	5(56)	56	8(2)	25	4(46)	65	8(2)	22
Adductor magnus [Lyons et al. (1983),Winter et al. (1987),Olree et al. (1995), Hof et al. (2002)]	4(40)	66	27(33)	40	3(30)	72	29(38)	41
Medial hamstring [Ericson et al. (1986), Winter et al. (1987), Ounpuu et al. (1989), Nymark et al. (2005)]	4(39)	67	94(2)	28	-	-	-	-
Lateral hamstring [Olree et al. (1995), Winter et al. (1987), Yang et al. (1984)]	3(48)	54	98(7)	26	2(28)	57	3(0)	25
Semitendinosus [Pierotti et al. (1991), Hof et al. (2002),Den et al. (2004), Prosser et al. (2011)]	3(39)	76	91(2)	25	-	-	-	-
Bicep femoris [Bovi et al. (2011), Den et al. (2004), Ericson et al. (1986)]	3 (39)	54	99(5)	21	2(19)	65	3(9)	-
Vastus lateralis [Yang et al. (1985),Ericson et al. (1986), Winter et al. (1987), Ounpuu et al. (1989), Ciccotti et al. (1994), Hof et al. (2002), Nene et al. (2004),Nymark et al. (2005),Chleboun et al. (2007),Barr et al. (2010)]	10(127)	65	6(4)	21	9(105)	66	6(3)	21
Vastus medialis [Ericson et al. (1986), Ciccotti et al. (1994), Hof et al. (2002), Den et al. (2004), Bovi et al. (2011)]	5(70)	53	3(3)	39	3(28)	70	6(0)	-
Rectus femoris [Yang et al. (1985),Arsenault et al. (1986), Ericson et al. (1986), Winter et al. (1987), Ounpuu et al. (1989), Pierotti et al. (1991),Ciccotti et al. (1994), Olree et al. (1995), Hof et al. (2002), Den et al. (2004),Nene et al. (2004),Nymark et al. (2005), Barr et al. (2010),Bovi et al. (2011), Prosser et al. (2011)]	15(183)	62	13(18)	35	12(131)	69	16(20)	34
Lataral gastrocnemius []Clancy et al. (2004),Ericson et al. (1986), Olree et al. (1995), Hof et al. (2002),Winter et al. (1987)]	5(54)	61	43(2)	24	4(44)	61	44(2)	26
Medial gastrocnemius [Ericson et al. (1986),Winter et al. (1987), Ciccotti et al. (1994),Hof et al. (2002), Den et al. (2004),Warren et al. (2004),Nymark et al. (2005),Murley et al. (2009),Bovi et al. (2011), Murley et al, (2014)]	10(171)	66	42(3)	25	8(129)	73	42(3)	25
Soleus [Yang et al. (1985),Arsenault et al. (1986), Ericson et al. (1986), Winter et al. (1987), Ounpuu et al. (1989), Ciccotti et al. (1994), Hof et al. (2002), Den et al. (2004), Bovi et al. (2011)]	9(119)	54	37(8)	21	7(77)	58	46(3)	16
Tibialis anterior [Yang et al. (1985),Arsenault et al. (1986), Ericson et al. (1986), Winter et al. (1987), Ciccotti et al. (1994), Olree et al. (1995), Hof et al. (2002), Den et al. (2004) Warren et al. (2004), Nymark et al. (2005), Chleboun et al. (2007), Murley et al. (2009), Bovi et al. (2011)),Murley et al, (2014)]	14(201)	64	100(9)	27	11(149)	69	100(10)	24
Tibialis posterior [Murley et al. (2009),Murley et al, (2014)]	2(42)	81	8(3)	30	-	-	-	-
Peroneus longus [Winter et al. (1987), Hof et al. (2002), Den et al. (2004), Murley et al. (2009), Bovi et al. (2011), Murley et al, (2014)]	6 (92)	69	39(15)	48	5(72)	76	39(17)	49

Table 3.4 Summary table for muscles and average standard deviation of the grand EMG profiles in group 1 (ascending order of average standard deviation)

Muscles	Average standard deviation across the gait cycle (%)
Without studies with QS < 50%	
Soleus	16
Vastus lateralis	21
Gluteus medius	22
Tibialis anterior	24
Lateral hamstring	25
Medial gastrocnemius	25
Lateral gastrocnemius	26
Gluteus maximus	28
Rectus femoris	34
Addustor magnus	41
Peroneus longus	49
With all included studies	
Bicep femoris	21
Semitendinosus	25
Medial hamstring	28
Tibialis posterior	30
Vastus medialis	39

Table 3.5 Analysis table for reported muscle Group 2 and 3

Muscles (Studies)	Studies (N)	Average QS (%)	Timing of Peak (%)	Average SD (%)
Group 2 Muscles reported by a single study with between-subject variability				
Gluteus minimus				
Anterior compartment [Semciw et al. (2014)]	1(15)	76	41	20
Posterior compartment [Semciw et al. (2014)]	1(15)	76	13	16
Adductor longus [Winter et al. (1987)]	1(16)	61	66	49
Sartorius [Winter et al. (1987)]	1(15)	61	72	46
Extensor digitorum longus [Winter et al. (1987)	1(11)	61	8	23
Group 3 Muscles reported without between-subject variability				
Gluteus maximus				
Upper compartment [Lyons et al. (1983)]	1 (10)	79	5	-
Lower compartment [Lyons et al. (1983)]	1 (11)	79	4	-
Gluteus medius				
Anterior compartment [Semciw et al.(2013)]	1(15)	76	13	-
Posterior compartment	1(15)	76	12	-
Middle compartment	1(15)	76	12	-
Tensor fascia latae [Lyons et al. (1983)]	1(9)	79	37	-
Bicep longhead [Lyons et al. (1983),Ciccotti et al. (1994), Hof et al. (2002)]	3(42)	60	94	-
Semimenbranosus [Lyons et al. (1983),Ciccotti et al. (1994), Hof et al. (2002)]	3(41)	59	92	-
Popliteus [Davis et al. (1995)]	1(9)	66	94	-

Grand EMG profiles

EMG profiles were obtained by digitising figures from 24 studies involving a total of 345 participants and 30 muscles with average sample size of 87 (Table 3.2). Given that excluding studies with lower quality scores led to a decrease in the variability of data for several clinically important muscles only data from studies with a quality of score of greater than 50% were pooled to form grand EMG profiles. Figures 3.2 and 3.3 thus illustrate 16 grand EMG profiles synthesized from 20 studies for lower limbs from Group 1. Figure 3.2 shows traces from all included studies regardless of the quality scores. This allowed direct comparison between mean traces from different studies regardless of the number of participants. Figure 3.3 shows the grand EMG profiles pooled from studies with high quality reporting (over 50% quality score). One and two standard deviation bars plotted in different shades of grey reflected both between-subjects variability and variability in techniques between pooled studies. These EMG profiles had an averaged quality score of 68% and have an averaged SD of 28% of peak magnitude. They were pooled from 263 participants (age 15-58 years) walking at normal/self-selected averaged speed 1.1 m/s. It can be seen that variability (represented by the width of the SD) varies considerably from muscle to muscle. Most data on muscles in this group were robust based on high quality reporting, with groups of participants and reports of between-subjects variability. Two muscles peroneus longus and adductor magnus were particularly variable with over 40% SD. While studies with over 50% quality score did not report vastus lateralis and vastus medialis with between-subjects variability (Figure 3.3).

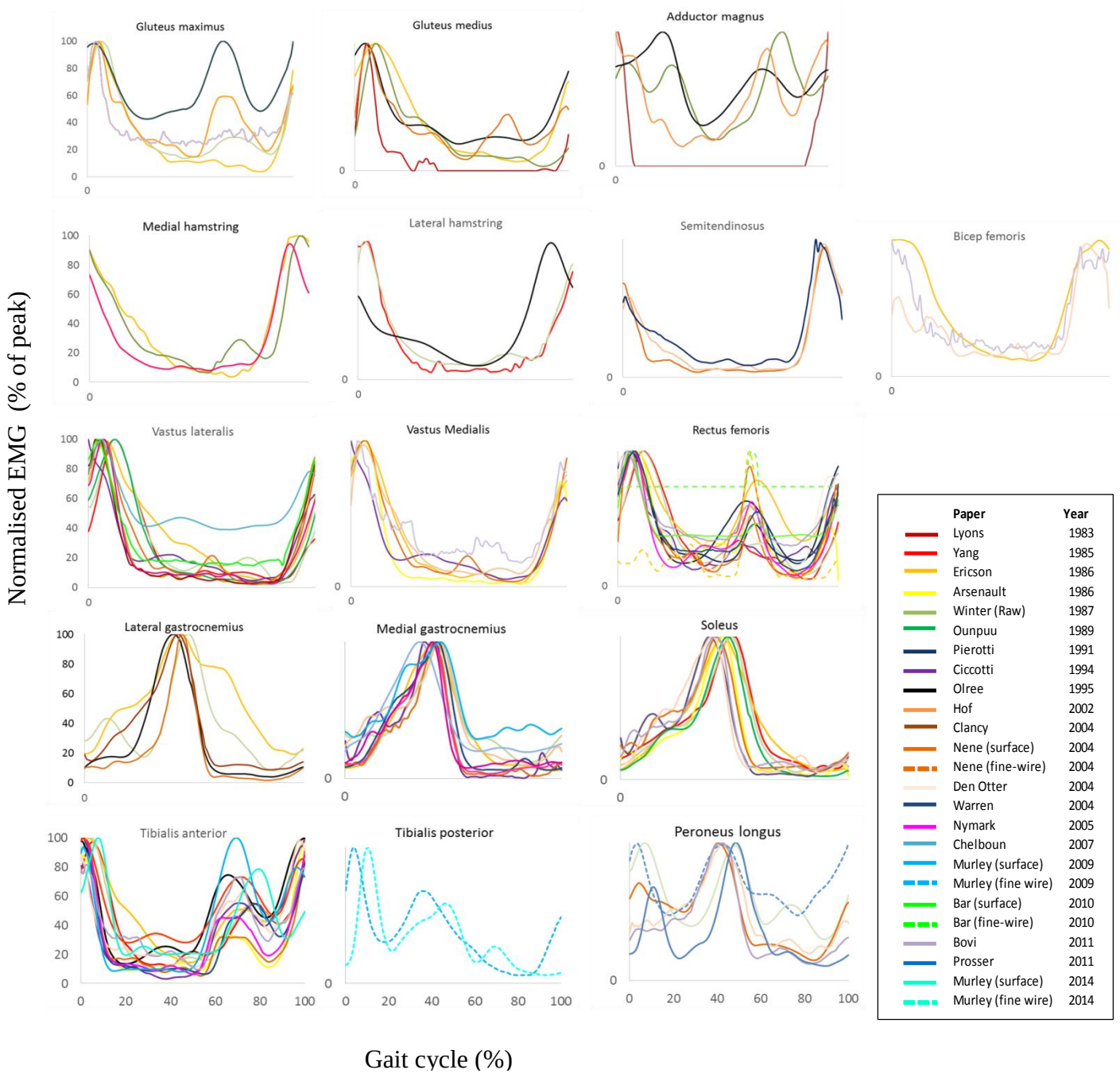


Figure 3.2 EMG profiles reported by multiple studies with standard deviation (Group1)

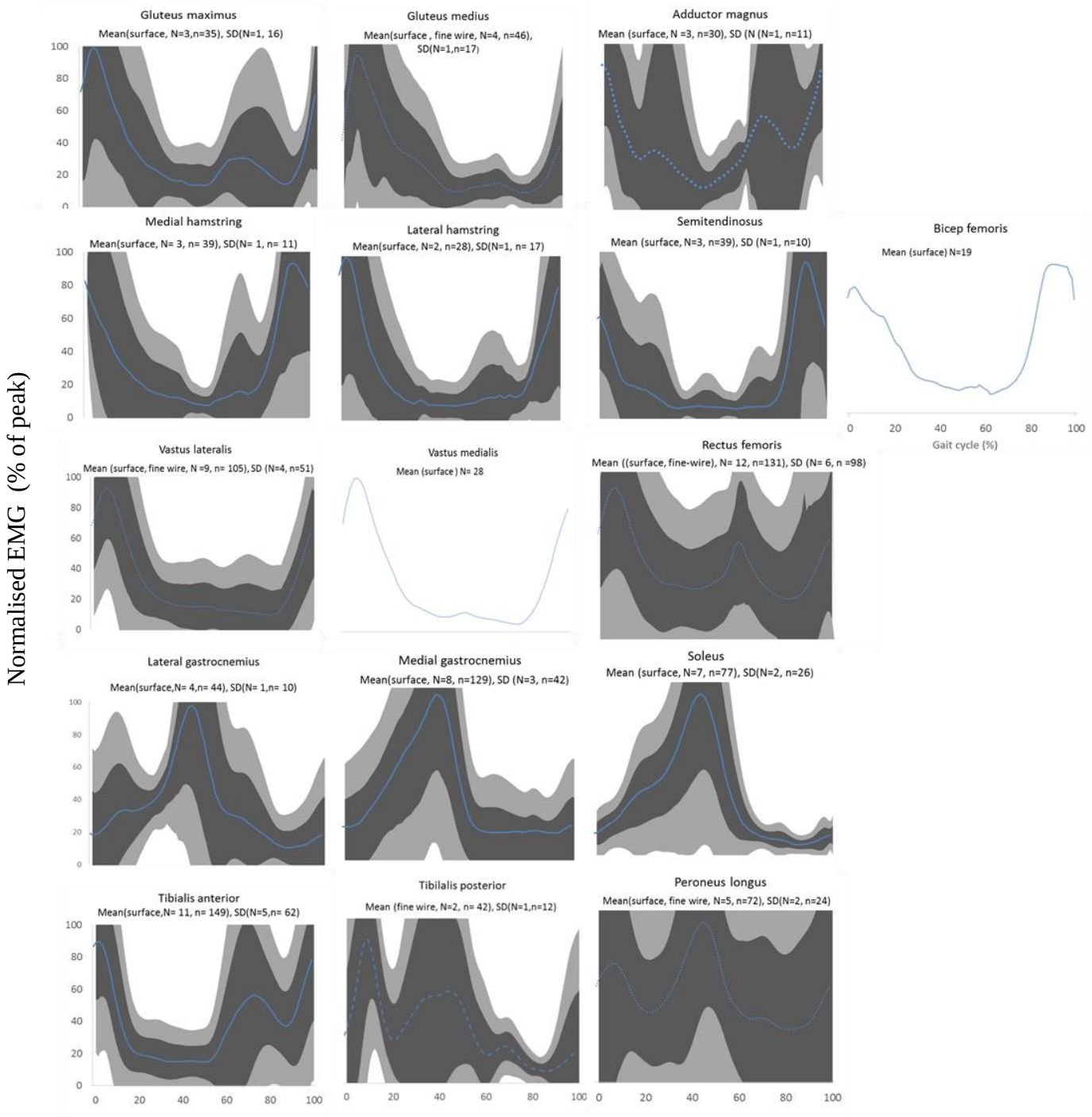


Figure 3.3 Group 1 Grand EMG profiles (N=number of included paper, n= number of participants, dark grey=±1SD and light grey = ±2SD)

Group 2: muscles reported by a single study with between-subjects variability

For 5 muscles (gluteus minimus: anterior and posterior compartments, extensor digitorum longus, sartorius, and adductor longus), only one study was identified with between-subjects variability (Figure 3.4). The averaged sample size for these muscles was 14 with the average quality score of 67% and the average SD was 31%.

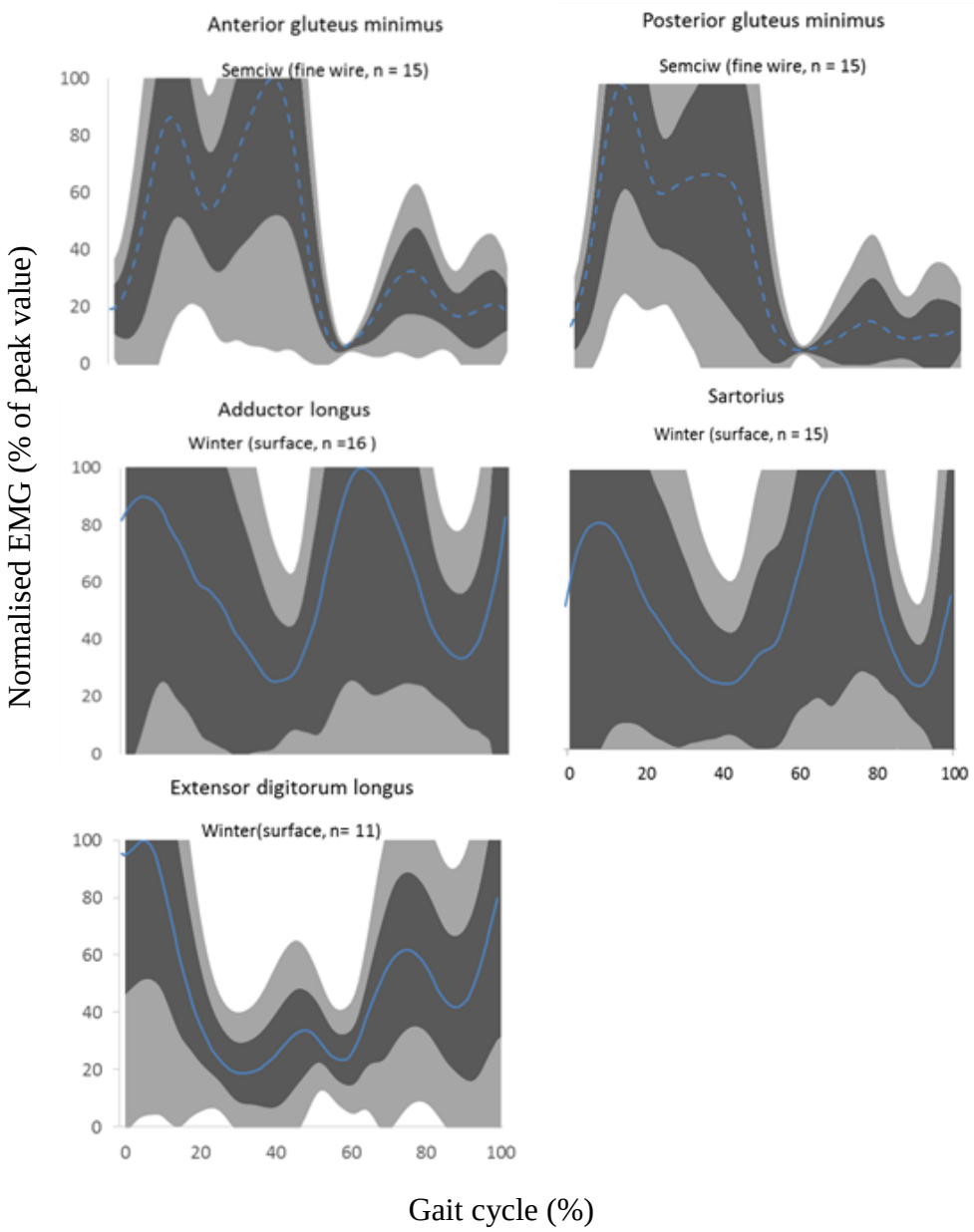


Figure 3.4 Group 2 Grand EMG profiles (n= number of participants)

Group 3: muscles reported without intersubject variability

There were 9 muscles (upper and lower compartments of gluteus maximus, anterior, middle and posterior compartments of gluteus medius, semimembranosus, biceps femoris long head, popliteus and tensor fascia latae) for which data had only ever been reported as mean value. All of them were measured using fine-wire sensors, while biceps femoris long head and semimembranosus were measured by both types of sensors. The averaged sample size was 19 with averaged quality scores of 72%. None of these reported the between-subjects variability. (Figure 3.5)

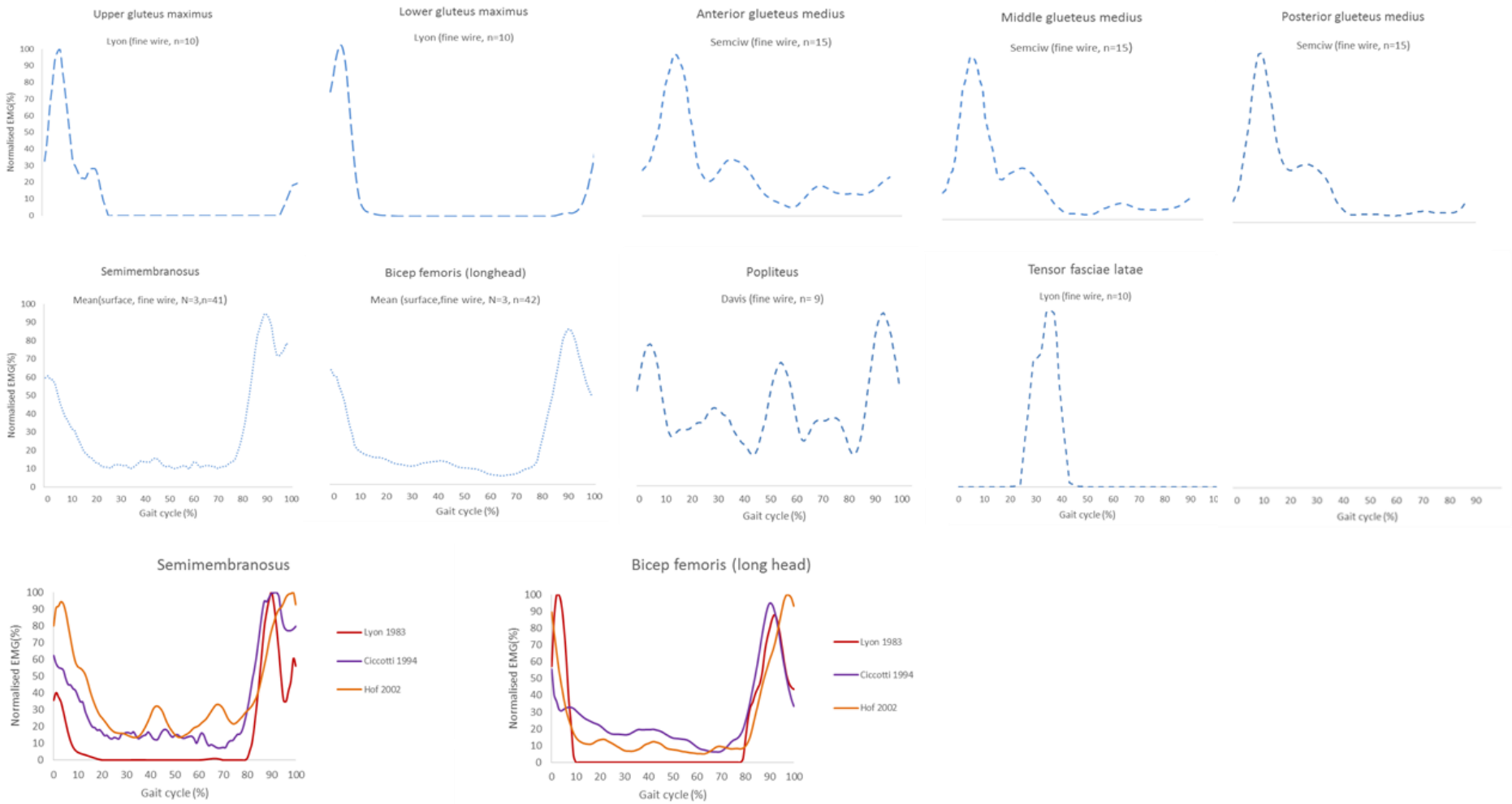


Figure 3.5 Group 3 Grand EMG profiles

3.8 Discussion

3.8.1 Study characteristics

This was the first systematic review to determine the consensus in the evidence base for normative lower limb EMG profiles across the gait cycle in healthy adults. This is important because understanding normative EMG profiles is an essential prerequisite for clinical interpretation of gait analysis data in patient groups. Perhaps the most surprising overall finding of this review was just how few studies report EMG activity of lower limb muscles during straight walking in healthy young adults in the peer-reviewed literature. For most muscles, the number of studies and sample size were small leading to the need for this systematic review. It was only possible in five muscles (vastus lateralis, rectus femoris, medial gastrocnemius, soleus, tibialis anterior) to construct EMG profiles from more than a hundred participants, and these required data from a minimum of eight studies (Figure 3.2-3).

The quality of reporting was quite variable (range from 42% to 86%) (Table 3.2). Several guidelines and recommendations for both types of sensor placements were used across identified studies: Chapman et al. (2006), Perotto et al. (2005), Leis et al. (2000), SENIAM (Hermens & Merletti, 1999), Perotto (1994), and Zipp (1982). Many papers simply contained a description of the locations (Arsenault, Winter, & Marteniuk, 1986; Chleboun, Busic, Graham, & Stuckey, 2007; Clancy, Cairns, Riley, Meister, & Kerrigan, 2004; Davis, Newsam, & Perry, 1995; Lyons et al., 1983; Olree & Vaughan, 1995; Warren, Maher, & Higbie, 2004; Winter & Yack, 1987; Yang & Winter, 1985), which were similar to the SENIAM recommendation and so most studies score fully for this aspect of collection. The crosstalk tests and description of data acquisition were frequently inadequate to assure

quality of collection. The quality score only reflected quality of reporting the methodology which, of course, does not necessarily reflect on the actual quality of the research itself.

3.8.2 Synthesised grand EMG profiles and standard deviations

The quality of data collection is important to ensure reliable EMG profiles and this is supported by the decrease in variability seen when only studies with the high reporting quality included. The variation indicated by SD grey area is large during active periods of EMG and largest at the time of the peak (Figure 3.3). Average SDs (%) calculated in this review consists of both between-subject and between-study variability in EMG measurements. When we excluded studies with quality score <50%, the SD reduced considerably in most muscles suggesting improved agreement between studies with higher quality assurances in collection methods. With the studies with over 50% quality score, the average SD (%) reported in this review is broadly similar to that previously reported for joint moments (Winter, 1991). From Winter's data (Winter, 1991), the average SD as a proportion of the peak signals were calculated as an average across the gait cycle to be 9%, 31% and 41% for ankle, knee and hip moments respectively. These values are directly comparable to the average SDs for EMG activity as a proportion of peak signal presented in Table 3.3. For the hip and knee muscles the variability in EMG (range of 21-41%) is similar to that for joint moments. However, the average SDs from EMG of ankle muscles (16-48%) are much larger than that of the ankle moment (particularly for peroneus longus). The SD of these kinetic graphs from Winter's study represented the between-subject variability as the data acquisition and processing are identical between subjects. Therefore, EMG profiles for some muscles at the hip and knee are as reliable as the kinetic data of hip, knee and ankle which are widely used in CGA. These are a little higher than SD reported in kinetic measures but can be much higher for some. The SD of these EMG profiles may be overestimated for

the between-subject variability as the variability in technique between studies will also have contributed to it. However, the SD are close to those of kinetics allowing some confidence in our synthesized profiles as a reference: provided that gait analysts followed the recent guidelines for EMG collection and processing to ensure reliable signals which are comparable to the synthesized profiles in this review.

In this review, the synthesized EMG profiles would be considered along with the kinematics which are also commonly reported in CGA. The following section describes how synthesized profiles of muscle activity derived from studies with high quality report relate to biomechanics of gait cycle. This should allow the confidence in consensus of the data to be compared with the local gait EMG as part of quality assurance. This process has been recommended for kinematics and kinetics (Baker, 2006; Pinzone et al., 2014).

Ankle

The research papers pooled to create the grand EMG profiles of tricep surae (soleus, medial and lateral gastrocnemius) were of lower quality (averaged quality score 64% compared to average of 66% across all studies) but showed lower between-subject variability (SD of 22% compared to average of 28%) (Figure 3.3, Table 3.3). This suggested that although there were differences in collection methods, the reports on muscle activity agreed. Muscle activity increased through stance to reach a peak around the time of opposite foot contact (50% gait cycle). During this time the muscles controlled tibial advancement during forward progression throughout stance and contributed to heel rise to support body weight in the second half of stance (Perry, 1992). The general shape of the EMG profile through stance is broadly similar to that of the internal plantarflexing moment recorded using inverse dynamics.

Activity in the major dorsiflexor (tibialis anterior) was reported from a range of studies with similar quality scores (average quality score 60%) and between-subject variability (SD of

24%) (Figure 3.3, Table 3.3). There was agreement between different studies that the muscle was most active in very late swing to avoid foot drop and continued into early stance when it was required to oppose the external plantarflexing moment provided by the ground reaction passing behind the ankle (Shiavi, 1985). The variability of EMG amplitude in early swing between studies was larger than expected. Some studies measured high levels of activity at this time whilst others measured much smaller levels despite the widespread assumption that the dorsiflexors were essential to provide dorsiflexion at this time to achieve foot clearance (Winter, 1991).

Activity of tibialis posterior was reported by the same group of researchers in 2009 and 2014 (Murley, Buldt, et al., 2009; Murley, Menz, & Landorf, 2014) (Figure 3.3, Table 3.3) . Both reports had high quality scores compared with the average across all included papers resulting in the highest averaged quality score (81%) across all reported muscles in this review. Both reported peak activity in early stance which was consistent across individuals. There was much more variability between individuals across the second half of stance. This may be a result of the high variability in foot posture or physical characteristics as demonstrated by a sub-analysis of the first study (Murley, Buldt, et al., 2009) (despite that study only recruiting individuals with a static "normal" foot posture). The simultaneous ankle kinematics and kinetics may be useful for better understanding of tibialis posterior functions and identification of impairments for clinical gait analysis as it played the important role in foot posture in patients with neuromuscular disorders such as stroke and cerebral palsy.

There was large variability compared with the mean variability across all muscles in peroneus longus profiles reported in five papers from good sample size (n=92) (Figure 3.3, Table 3.3). This appeared to come from variability both within (Murley, Buldt, et al., 2009; Winter & Yack, 1987) and between studies. Murley, Buldt, et al. (2009) illustrated different

patterns of peroneus longus from different participants suggesting that the variability was a result of different foot postures. Activity of the muscle in mid-single support was greater in participants with a normal arch than it was in those walking with flat feet. The high inconsistency in the profile (SD =49%) suggested that further work is required before clinical use of EMG measurements from this muscle.

In summary, among the muscles acting across the ankle, triceps surae and tibialis anterior were considered as exhibiting sufficiently low variability making them suitable for clinical use. However, tibialis posterior and peroneus longus show much larger variability and further work is required to understand this before clinical use is possible.

Knee

Only two of vasti had EMG profiles reported in this review (Table 3.3). Both vastus lateralis and vastus medialis were reported with sample sizes n = 105 and 70 respectively and the variability was below average (21% compared to 28%). In Figure 3-2 and 3-3, the included studies agreed upon the activity at early stance to control knee flexion and assist knee extension Hof et al. (2002), Bovi, Rabuffetti, Mazzoleni, and Ferrarin (2011) and Ericson, Nisell, and Ekholm (1986) reported small bursts of activity in these muscles in late stance or early swing which were not presented in the other studies.

EMG profile of rectus femoris showed larger variability than most muscles (SD = 34%) being derived from a large sample (131 participants) from 12 studies with averaged quality score 69% (Figure 3.3, Table 3.3). The grand profile showed two bursts of activity. The first, in first double support, can be associated with the knee extensor moment resisting flexion and then causing extension. The second, around foot off, may help accelerate the hip into flexion and control the acceleration of the knee into flexion (Winter, 1991). The fine-wire EMG profiles from Barr et al (2010) and Nene et al (Nene et al., 2004) showed rectus femoris activity during transition from stance to swing only and illustrated that the other burst of

activity was crosstalk from vastus lateralis. The profile from Ciccotti et al (Ciccotti, Keran, Perry, & Pink, 1994) obtained from fine-wire sensors, however, had a similar pattern to that from the surface sensors. The high variability associated with this muscles and concerns over whether the surface sensor signal accurately represents muscle activity should lead to clinical measurement being treated with some caution.

There was some ambiguity in how papers refer to the hamstrings either as muscle groups or as individual muscles and we thus presented data both for the medial and lateral groups and for the two main muscles of each group (semitendonosus, and biceps femoris) following the terminology used by different authors (Table 3.3). It can be seen from figure 3.2 and 3.3 that the recorded EMG activity was reasonably similar across all the hamstrings. The average quality of muscle profiles in this group was 66% with the average sample size of 31 and the average SD of 25% below the average SD across all muscles. The pooled EMG profiles of these muscle groups showed agreement on a single burst of activity during transition between swing to stance. This is responsible for slowing knee extension through most of late swing, initiating knee flexion in very late swing and contributing to the hip extensor moment in early stance (Shiavi, 1985; Winter, 1991). The timings of the peaks were in agreement with Winter's suggestion that medial hamstrings may serve primarily as a knee flexor while lateral hamstrings mainly served as a hip extensor (Winter, 1991) (although this was not apparent in data from the individual muscles, semitendonosus, and biceps femoris). Moreover, some participants showed small activity of medial hamstrings in early swing. Between-subject variability was particularly large during active phases for muscles in this group.

The activity of muscles mainly acting on the knee were frequently measured resulting in a high number of included participants to form the normative database. The grand EMG profiles of medial hamstrings, lateral hamstrings, semimembranosus and vastus lateralis

showed good agreement between studies and subjects giving high confidence in clinical studies to identify abnormality. However, there were insufficient reports for biceps femoris (as a single muscle) and vastus medialis from high quality report to have confidence in any particular pattern. The main concern with rectus femoris is the large SD was the EMG recorded from different sensor types. Therefore the EMG measurement of rectus femoris and clinical application (3.2 and 3.3) should be carried out with some caution.

Hip

Gluteus maximus profile was pooled from a reasonable sample of 35 participants using surface sensors with averaged quality score 65% less than the average score across all muscles and averaged SD 28% similar to the average SD of all muscles (Figure 3.3, Table 3.3). All three included studies agreed on the peak occurring in the middle of 1st double support to which would be for concentric hip extension (Bovi et al., 2011; Hof et al., 2002; Shiavi, 1985; Winter & Yack, 1987). They also showed a smaller burst of activity in early swing in some people, similar to medial hamstrings. Winter et al. (1991) suggestion that this was to control the forward swing of the thigh is not particularly convincing, but there is no other obvious explanation. Though the patterns were similar, different studies showed different amplitudes. This difference may be caused by individuals' characteristics as similar variation appears in between-subjects variation obtained from Winter's study(1987).

The EMG profile of gluteus medius, a hip abductor, was pooled from a sample (n=46) and an averaged quality score 65% and average SD 22% below the average of all muscles (Figure 3.3, Table 3.3). The grand mean from four studies showed peak of activity in the middle of first double support which fell away rapidly through single support. It was interesting that this was earlier and shorter than would be predicted if the main function of this muscle was to provide the substantial hip abductor moments required to support body weight during single support. The coronal plane hip abductor moment is consistently recorded as having

a “double bump” pattern (reflecting the ground reaction) with peaks in early and late single support. Between studies, the timing of peak was similar but the length of active periods were slightly different (10-20% gait cycle). This may be caused by individuals’ characteristics as this difference was also seen in between-subject variability (SD). There was some evidence of a smaller burst in mid-swing (observed in other reports (Battye & Joseph, 1966; Shiavi, 1985)) which may assist foot clearance (Shiavi, 1985). However, this could be difference between sensors. Only means of surface EMG from three studies showed minor activity in early swing, the activity was prominent in Hof’s mean (Hof et al., 2002). Whereas mean of fine- wire EMG from Lyons ‘study showed absence of activity in swing (Lyons et al., 1983).

The glutei are large muscles and it has been suggested that there is anatomical and functional justification for considering them as having separate compartments (Perry, 1992). Several studies placed fine-wires in the different compartments (Lyons et al., 1983; Semciw, Pizzari, Murley, & Green, 2013), three for gluteus medius and two for gluteus maximus. Although the authors of these studies drew attention to perceived differences in the data the overall appearance was of a pattern of broadly similar activity well represented by the grand mean depicted in Figure 3.5. These studies reported mean signals without any indication of variability so it was difficult to assess what evidence there was for differences between the activation of the different compartments.

Adductor magnus profile, a hip adductor and extensor, was pooled from three studies with 30 participants and averaged quality score of 72% (Figure 3.3, Table 3.3). The grand mean had a peak of activity at foot contact suggesting some role in stabilizing the hip at the transition between swing to stance (Lyons et al., 1983; Winter & Yack, 1987). At other places in the gait cycle there was much less agreement between studies leading to poor overall variability (SD = 41%) and between participants (Olree & Vaughan, 1995; Winter

& Yack, 1987). This may cause difficulty in identification of pathological feature in CGA. Although the authors of many studies proposed interpretations of their own data (Basmajian & De Luca, 1985; Leighton, 2006; Lyons et al., 1983; Winter, 1991; Winter & Yack, 1987), the difference between the studies rendered these somewhat questionable. It maybe that the secondary transverse plane action of these muscles was important (Leighton, 2006; Winter, 1991). The one fine-wire study of this muscle (Lyons et al., 1983) showed only a single burst of activity occurring around foot strike and it may be thus that activity elsewhere detected by surface sensors was a cross-talk artefact from adductor longus, sartorius or medial hamstring. The use of kinematics and kinetics may lead to better understanding of the muscle functions during gait.

The grand EMG patterns of muscles acting across the hip showed good agreement between studies and subjects allowing for identification of abnormality, except in adductor magnus (Figure 3.3). However, the between-subject variability of muscles in this group were large as a result of different amplitudes despite the scaling using their peak value. This aspect suggested the need for better understanding of their functions in individuals using other data such as kinematics, kinetics or centre of pressure trajectories. The use of fine-wire sensors allowed different compartments of gluti to be measured. Therefore their EMG measurement and clinical application should be carried out with some caution. Also, different sensors yielded different EMG profiles of adductor magnus and with the large SD suggested that the further study was required.

Summary

To summarise, there is large between-subject variability in measured EMG signals for all muscles ranging (amongst muscles studied) from 16% to 49% of peak value (Table 3.4). Given that, by definition, 35% of data from any healthy individuals falls outside the +/- 1 SD range, signals recorded clinically will have to be markedly different from the mean

normative trace to give confidence that muscles are functioning abnormally. Considerable caution will be required in making clinical recommendations on the basis of EMG alone. It may be, however, that data from the more repeatable muscles is strong enough to augment data from kinematic and kinetic measurements.

Table 3.4 ranks the muscles from least to most variable. Two muscles, adductor magnus and peroneus longus exhibit substantially more variability than the others. It is difficult to see how clinical EMG s ignals from these muscles could be used to identify abnormality against this data. The other muscles SD values spanning a range from 16% (soleus) which might be considered as quite reasonable for clinical use to 34% (rectus femoris) which is much more questionable. There does not seem any particular reason for suggesting a specific threshold value to indicate which EMG signals are clinically useful and which are not but Table 3.4 has value in allowing clinicians to have some idea of which muscles show more and which less consistency within the healthy adult population. Most of the muscles assessed most commonly (gastrocnemius, soleus, vasti, hamstrings) are amongst the most consistent giving reasonable justification for clinical use. Rectus femoris is the other commonly assessed muscle but its high variability (34%) suggests particular caution in concluding abnormality.

3.8.3 Identify issues for further investigations

In addition to tibialis posterior, peroneus longus, biceps femoris, vastus medialis and adductor magnus in Group 1 (Figure 3.3), muscles in Group 2 (Figure 3.4) and 3 (Figure 3.5) demonstrated a lack or inadequate data sets for a consensus to be used as normative database with between-subject variability and a need for further investigations. Although muscles in group 2 were reported with the between-subject variability (averaged SD 31%), they were reported by a single group of study with relatively small averaged sample size (n= 14) and good quality score (67%). 9 muscles in group 3 were reported without between-

subject variability reported in the same study. Their quality scores were relatively high with average of 72% detected from a small averaged sample size (n= 19). Without between-subjects variability, the uses of normal EMG profiles were limited. Thus muscles group 2 and 3 should be further investigated.

30 muscles were reported in this review. The selection for studies may be based on the perceived importance in gait analysis and practicality in EMG measurements. For example of intrinsic foot muscles playing roles in foot posture during gait, there was a lack of reports of EMG profiles due to small size and difficulty in measurement. There were also lower limb muscles which were not reported in any of the included studies such as psoas, even though it was one of targeted muscles for correction of crouch gait (Delp, Arnold, Speers, & Moore, 1996).

Out of 30, only 10 profiles in Group 1 (Figure 3.2-3) had a good consensus as a normative data for clinical use and they were superficial muscles. The recent publications of the EMG database such as Schwartz et al. (2008) provided gait analysis data: spatial temporal parameters, kinetics and kinetics but the EMG profiles were measured from children and showed only muscle activity of the superficial muscles. They also reported these parameters across different speeds which had effects on walking patterns and muscle recruitments (Den Otter et al., 2004). Hof et al. (2002) proposed the functional groups to estimate the EMG for each muscle at any speeds from a constant and a proportionally increasing factor. Several attempts were made to establish the typical profile when the participants walk with a very slow speed which was frequently seen in the patients with upper-motor-neuron deficits but they were limited to superficial muscles (Den Otter et al., 2004; Nymark, Balmer, Melis, Lemaire, & Millar, 2005; Schwartz et al., 2008). Therefore there is still a need to develop a normative database consisting of EMG measurements on clinically important muscles

regardless of using surface or fine-wire sensors with other gait data across different speeds for use in clinical applications.

As a result it was found that the activity of superficial muscles were investigated in the majority of papers but for the deep muscles such as tibialis posterior, little normative reference existed. Tibialis posterior is potentially important in foot and gait biomechanics as it is a muscle that holds the medial longitudinal arch of the foot to ensure appropriate position of the ground reaction. It also acts as inverter, and plantarflexor of the foot. It likely contributs to foot-drop and equinovarus foot posture in patients with post–stroke. In patients with cerebral palsy and post-stroke, this muscle is one of those targeted for botulinum toxin. So the measurement of tibialis posterior activity should be included in clinical gait analysis for these patient groups.

Muscles act to exert moments about joints and this should lead to some correspondence between muscle activity and joint kinetics. Whilst for many muscles such a correspondence is evident this there are exceptions. For example, gluteus medius is generally accepted to be the primary hip abductor yet it is only recorded as being active in early stance whereas the hip abductor moment is substantial throughout stance. In many individuals there is also evidence of gluteus medius activity in early swing which does not correlate with joint moment data. Further work is required to understand in detail how activity in some specific muscles relates to production of joint moments.

3.9 Limitations

As with any studies, there were a few limitations in this review: the most obvious are those associated with pooling data from heterogeneous study methodologies which is often exacerbated by the limited information describing these in the articles. Firstly, to determine whether there was a consensus of EMG profiles, a number of studies were pooled to form a grand EMG. To overcome this, the inclusion and exclusion criteria strictly complied with

the available EMG standard and guidelines to limit the heterogeneity. The digitized EMG profiles were also scaled/normalised by their peak value to transform data to the same unit (%) before pooling. The preliminary analysis showed similar EMG profiles of tibialis anterior regardless of different normalisations, walking on platform and treadmill. Furthermore, the average SD reported in Group 1 was similar to those reported in well-established kinetics from a single study without signal variability due to employed different methods. Therefore, the EMG profiles could be as reliable as the kinematics in CGA.

Secondly, the review was based on information provided in the journal articles and supplementary materials only. The graphic resolutions obtained from the studies were varied and this may affect the quality of the digitised graphs and ultimately the EMG profiles presented in this review. Additionally, in Lyons et al. (1983) the digitized linear curve was obtained from histogram of integrated EMG profile starting from swing phase to stance phase. This transformation of data may cause variation. So interpolation and smoothing techniques were used to ensure the digitised graphs were similar to the original graphs in the articles before data pooling.

3.10 Conclusion and recommendations

There is large between-subjects variability in measured EMG signals for all muscles ranging (amongst the muscles studied) from 16% to 49% of peak value. Therefore, considerable caution should be applied when using EMG profiles alone in making clinical decision. Simultaneous kinematics and kinetics should be useful. Moreover, in order to obtain the consistent data, the EMG recommendations should be followed.

A number of EMG profiles of superficial muscles (gastrocnemius, soleus, vasti and hamstrings) are more repeatable than the others and serve as consensus for comparator in the quality assurance process. To support identification of muscle dysfunction, there is no specific threshold to indicate which EMG profiles are clinically useful. This review can

inform the clinicians how consistency of each muscles is within the healthy adult population and indicate the muscles requiring further investigation which could be helpful in CGA.

Chapter 4 Experimental procedure

Before clinically relevant information from EMG can be established, it is necessary to make a decision about how to collect EMG data. Specifically, the systematic review (Chapter 3) highlighted the need to resolve the issues of which normalisation approach to use, how best to collect and process fine-wire data, and that there was a lack of information on the tibialis posterior which could be of key clinical relevance. The tibialis posterior is one of the muscles that play an important role in the ankle-foot complex in patients with neurological disorders, e.g. stroke and cerebral palsy.

In order to address the paucity of information on the activation of the tibialis posterior during the gait cycle, certain methodological issues (the normalisation and processing of fine-wire data) must first be established before robust EMG studies of the tibialis posterior can be undertaken. This chapter describes the method for EMG data collection, which was used in all subsequent chapters. Chapter 5 aims to identify the most appropriate normalisation scheme. Chapter 6 explores how best to process data from fine-wire EMG in order for it to be comparable with established practices for surface EMG. Chapter 7 provides normative EMG dataset using these protocols. This experimental procedure chapter only outlines the common procedure and details about the sensors used; any differences to those described in this chapter are explained in detail in corresponding chapters.

This chapter covers details about the participants, EMG data collection, kinematic data collection, and data processing. This protocol will be used to: (1) compare two sets of fine-wire EMG signals detected from different locations in the same muscles (tibialis posterior, tibialis anterior and medial gastrocnemius) to test sensitivity to location; (2) compare the EMG signals detected by fine-wire and surface sensors in the same muscles between subjects and between sessions; (3) collect EMG signals during maximum voluntary isometric contraction (MVIC) to compare different normalisation schemes; and (4) collect

EMG signals from walking at different speeds in healthy individuals, both younger and older groups, to explore the differences between age groups for the formation of a definitive normative profiles.

4.1 Participants

This pilot study aimed to recruit two groups of healthy volunteers, one of younger volunteers (18-50 years of age) and the other of older healthy volunteers (over 50 years of age), to explore the differences between those age groups. A repeatability study was carried out in the younger group to evaluate different normalisations and sensor types. The older group data was collected for a normative dataset only. This is because there is a potential application in older pathological groups, such as stroke participants. With no prior data, it was not possible to provide a formal calculation of the necessary sample size. Similar studies comparing different electrode types have been carried out by Bogey et al. (2000) and Nene et al. (2004) who included five participants. A study on the tibialis posterior carried out by Murley, Buldt, et al. (2009) had 15 participants. So this study aimed to recruit ten participants which is a convenient sample size to explore these technical issues regarding fine-wire and surface EMG measurements.

The inclusion criteria were: healthy volunteers, self-reporting as free from any cardiovascular, musculoskeletal or, neurological injury or disease; not taking any anti-biotic or anti-coagulant medication or having anti-platelet therapy; having no immune deficiency conditions. Volunteers who were pregnant or had gait deviations or medical problems limiting exercise tolerance were excluded.

4.2 Procedure

4.2.1 Recruitment

An invitation letter with an attached poster and participant information sheet was circulated by e-mail, and recruitment posters were put on notice boards in the College of Health and Social Care at the University of Salford (Ethical approval number HSCR13-35). Potential volunteers then contacted the investigator directly by e-mail.

Four days prior to the scheduled experimental day, participants were advised to avoid any strenuous activity which might cause muscle soreness so as to ensure normal walking during data collection. Participants were asked to attend the testing sessions (scheduled at their convenience) wearing shorts and a T-shirts to allow the easy application of EMG sensors and motion-analysis markers.

When participants agreed to take part, appointments were made to attend the Gait Laboratory. Only participants in the younger group were asked to attend on two separate occasions with a minimum of two weeks apart. This time interval between testing was to avoid any intramuscular haematoma and disruption of muscle tissue (which affect the reliability of EMG between sessions) from fine-wire insertion to diminish (Paakkari & Mumenthaler, 1974). Each testing session took approximately three to four hours, including the time to place markers and electrodes.

The information that was requested at only the first appointment was:

a) Personal details: name, date of birth, gender, and a brief medical history to ensure compliance with the inclusion criteria.

b) Anthropometric data: height, mass, leg length, ankle width and knee width.

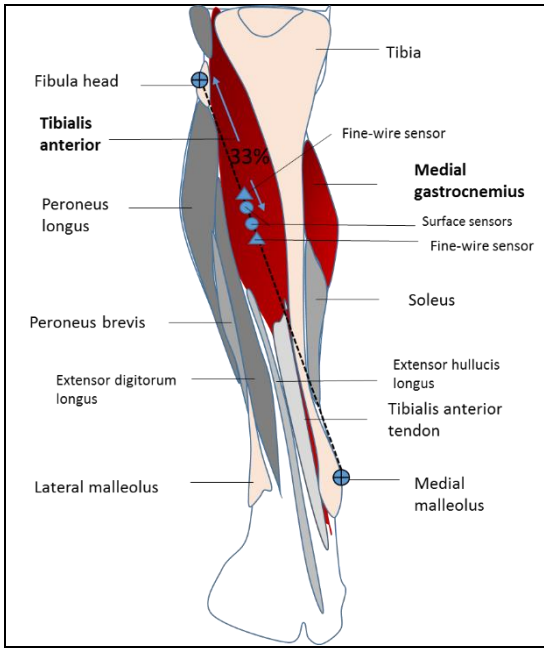
EMG, joint kinematic and kinetic data were recorded while walking during both appointments, as specified in the paragraphs below.

4.2.2 Electromyographic collection

EMG data were collected from the tibialis anterior and medial gastrocnemius using both fine-wire and surface sensors and from the tibialis posterior with fine-wire only. Two aims of this study are to investigate whether fine-wire EMG represents a signal of the entire muscle by looking at the variability of the EMG data detected by two fine-wire sensors located in the same muscles and the consistency of these with surface EMG data. This requirement guided electrode placement. Two fine-wire sensors were inserted 10 mm proximal and distal to the surface sensors on the tibialis anterior and medial gastrocnemius. Details about sensor placement are described in Tables 4.1-2.

For the tibialis posterior, two approaches for the insertion of fine-wire electrodes have been reported: anterior and posterior (Table 4.3). Guidance using ultrasound imaging under dynamic and non-weight bearing conditions is recommended (Murley, Buldt, et al., 2009). The anterior approach is half-way between the tibia and fibula through the interosseous membrane at the mid-third of the shank, with the posterior approach being halfway between the tibia tuberosity and the media malleolus (Murley, Buldt, et al., 2009). A posterior approach was the method used in this study, as it has been reported to have a lower dislocation rate during dynamic movement (Semple, Murley, Woodburn, & Turner, 2009). When using a posterior approach, there are two recommended positions relative to anatomical landmarks: halfway between the medial joint line of the knee and the medial malleolus (Chapman, Vicenzino, Blanch, Knox, & Hodges, 2010) and a more distal position halfway between the tibia tuberosity and the medial malleolus (Murley et al., 2009). Given that two placements were required for the protocol, both were used.

Table 4.1 Electrode placements for the tibialis anterior



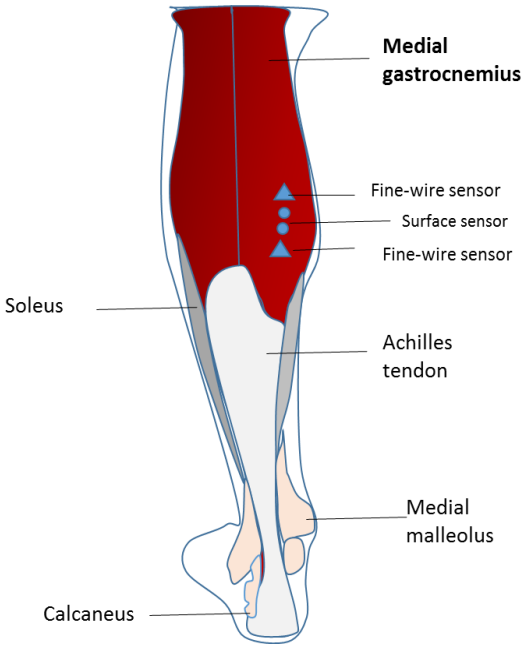
Location:
Approximately 33% of distance between the tip of the fibula and the tip of the medial malleolus (Hermens & Merletti, 1999)

Clinical test:
In a supine position with slight knee flexion 'support the leg just above the ankle joint with the ankle joint in dorsiflexion and the foot in inversion without extension of the great toes and apply pressure against the medial side, dorsal surface of the foot in the direction of plantar flexion of the ankle joint and eversion of the foot' (Hermens & Merletti, 1999).

Depth:
At 40% of muscle thickness (Chapman et al., 2010)

Note that the percentage for the depth of the needle is relative to the transverse plane muscle width seen on the ultrasound image.

Table 4.2 Electrode placements for the medial gastrocnemius



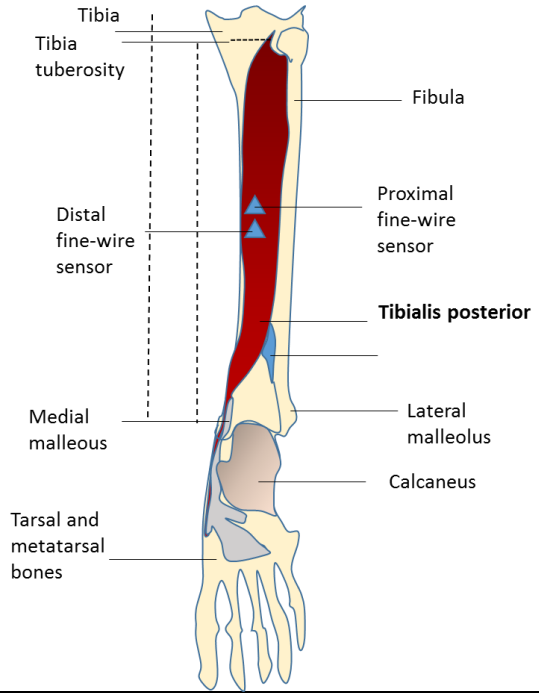
Location:
The most prominent bulge of the muscles (Hermens & Merletti, 1999).

Clinical test:
In a prone position with the foot projecting over the end of the table, 'plantar flexion of the foot with emphasis on pulling the heel upward more than pushing the forefoot downward. For maximum pressure in this position it is necessary to apply pressure against the forefoot as well as against the calcaneus'. (Hermens & Merletti, 1999)

Depth:
At approximately 60%, similar to the lateral gastrocnemius (Chapman et al., 2010)

Note that the percentage for the depth of the needle is relative to the transverse plane muscle width seen on the ultrasound image.

Table 4.3 Electrode placements for the tibialis posterior



Location:
Proximal sensor: 50% of the distance between the medial joint line of the knee and the medial malleolus (Chapman et al., 2010).
Distal sensor: 50% of the distance between the tibia tuberosity and the media malleolus (Murley, Buldt, et al., 2009)

Clinical test:
Use a muscle stimulator to induce inversion of the foot without flexion of the toes (Murley, Buldt, et al., 2009)

Depth:
At 50% of muscle thickness (Chapman et al., 2010)

Note that the percentage for the depth of the needle is relative to the transverse plane muscle width seen on the ultrasound image.

Surface EMG was measured using dual Noraxon EMG electrodes with disposable, self-adhesive Ag/AgCl, figure-of-8-shaped electrodes with dimension 40 mm x 22 mm. The diameter of each circular conductive area was 10 mm, and the inter-electrode distance was 17.5 mm (Figure 4.1). The application of the surface electrode followed SENIAM guidelines (Hermens & Merletti, 1999). First of all, an appropriate location for the electrodes was identified. The approximate location was as described in Table 4.1-3, with exact placement being guided by palpation of the muscle belly and bony prominences. A small area of the skin surrounding this was shaved (where necessary), rubbed lightly with abrasive gel to remove any dead skin, and then cleaned with a sterile wipe. Once the skin had dried completely, the electrode was placed. The sensors were then connected by short cables to a preamplifier/ transmitter unit which was attached to the skin with double-sided medical tape. An elastic bandage was wrapped around the surface electrode to ensure that nothing fell off during testing.

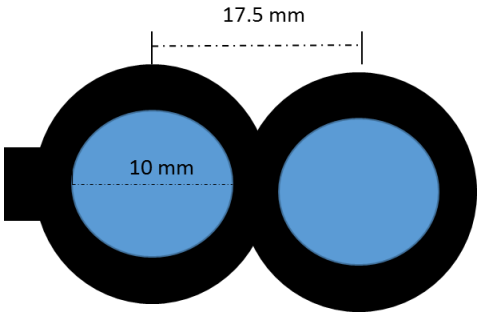


Figure 4.1 Dimensions of dual-surface electrode

The application of each of the dual fine-wire electrodes (44 gauge × 100 mm paired-hook wires, Teflon-coated stainless-steel wire) on the tibialis posterior, tibialis anterior and medial gastrocnemius involved the use of a sterile, unused hypodermic needle, 50 mm long and 25 gauge, to insert the wire pair into the muscle. Ultrasound imaging (MyLab70, Esaote, USA) was used to guide insertion using a 5–13 MHz linear array transducer (Barn, Rafferty, Turner, & Woodburn, 2012; Murley, Buldt, et al., 2009), as shown in figure 4.2. Once the electrode was inserted in the desired position, the needle was removed, leaving the wire electrodes in place. Then, the participants were asked to performed static isometric contractions to ensure the secured position of the fine-wire electrodes within the muscle (Rudroff, 2008). A small loop of wire was made before securing the electrode to the skin to allow for slight movement during dynamic contraction. A muscle simulator (Dantec Clavis, Natus Neurology Inc., USA) was then connected to the wire electrodes to lightly stimulate the muscle, thus causing a visible involuntary contraction to confirm correct positioning of the wire within the target muscle. Then, the wires were connected to a screw connector for each pair of fine-wire sensors. These fine-wire pre-amplifier connectors were then connected by short cables to a preamplifier/ transmitter unit which was adhered to the skin with double sided medical tape. The preamplifier/ transmitter units were identical for both surface and fine-wire sensors with an analogue output EMG gain of 2,000. Table 4-4, below, showed ultrasound images corresponding to the probe positions.

The participants described low discomfort (similar to insect bite) during the insertion of the needle and mild discomfort during the first few maximal contractions of the muscles and standing. After that they reported some sensation of the wire in the muscles but not discomfort. If similar tests are to be used clinically then it the likelihood of these sensations should be explained to patients in advance. Particular care may be required to reassure any patients, particularly children, who may have anxieties about any use of needles. In general it is felt that the mild discomfort will usually be justified by the potential benefits of the clinical information obtained using this technique.

The EMG signals were transferred to a laboratory processing computer via a wireless connection and a DTS desktop system (Noraxon U.S.A. Inc., Scottsdale, Arizona, U.S.A.). The sampling rate of the processing PC was 3,000 Hz with 16-bit resolution. The EMG pre-amplifiers had 1st order high-pass filters set to 10Hz +/- 10% cut-off. Under these conditions, the specified baseline noise was less that 1μV RMS, the input impedance was greater than 100 MΩ, and the common-mode rejection ratio was greater than 100dB.

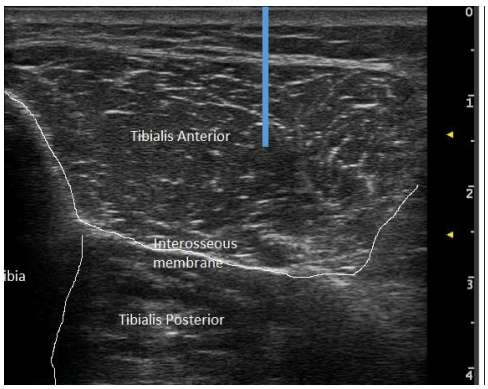
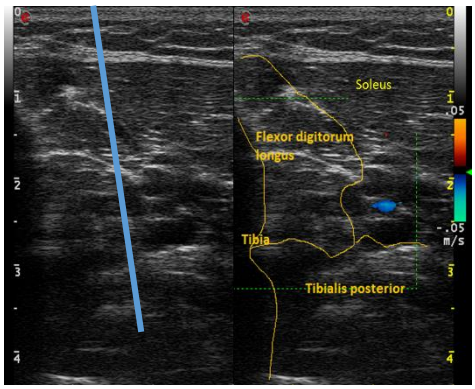
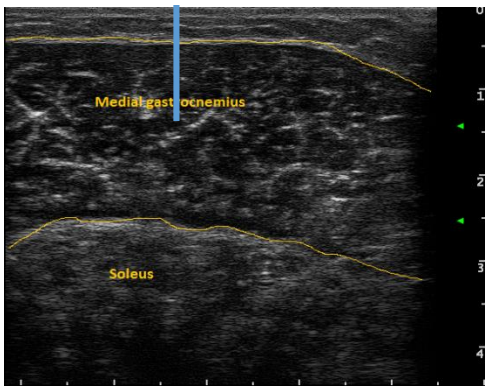
Probe position	Ultrasound image
	
	
	 Note the blue line is the direction and position of the fine-wire electrode

Figure 4.2 Examples of ultrasound images used to guide electrode placement (TA-tibialis anterior, TP-tibialis posterior and MG-medial gastrocnemius)

Figure 4.3 participant with electrode and sensors

4.2.3 Kinematic, kinetic and power analysis

Joint kinematic and kinetic data for the ankle, knee and hip were recorded to check that each subject 'walks are within the normal range' compared to Pinzone et al. (2014). 18 small reflective markers, each of 14 mm diameter, were placed on anatomical landmarks on the pelvis, thigh, shank and foot using double-sided tape in accordance with Plugin Gait with KAD model (Vicon, 2005) (Figure 4.3). The Gait Laboratory is equipped with an optoelectronic motion-analysis system with a Vicon software package, ten T-40 cameras (Vicon, Oxford, UK) (100 Hz) and four force plates (3,000 Hz) (Kistler Instrument Corp, Winterthur, Switzerland), with 20 N as the threshold. All data were synchronised in Vicon Nexus software (Vicon, Oxford, UK).

After completion of marker placement, participants were asked to walk along a 6 m walkway in the Gait Laboratory at five different speeds: a self-selected speed, 25% slower, 50% slower, 25% faster, and 50% faster. Walking speeds over the middle part of the walkway were measured using light gates–a test centre (TC) timing system (Brower timing system,

Utah, USA) consisting of two pairs of poles (2 m apart) with visual-beam emitters and sensors. Participants were given feedback on their speed of walking to allow them to achieve the target speeds. Participants were asked to walk six times at each speed (to a total of 30 walks) to retrieve a minimum of six successful gait cycles when the entire feet were on different force plates (Shiavi, Frigo, & Pedotti, 1998). Only, participants in the younger group repeated this testing session, approximately two weeks later, in order to investigate the between-sessions repeatability of measures.

A simpler protocol was required for the older participants. Tibialis posterior activity was measured using a fine-wire sensor as described in Chapman's guidelines (Chapman et al., 2010); tibilalis anterior and medial gastrocnemius activity was measured using surface sensors. The equipment and processing were identical to those for the young adults. Only one testing session was run for this group.

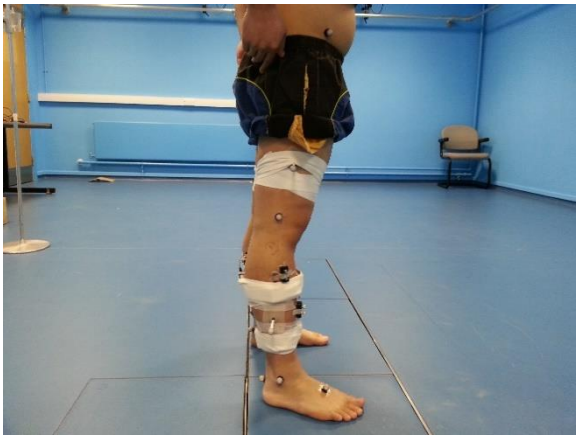


Figure 4.4 Participant with EMG sensors and markers

4.3 Data processing

The force platform data assisted defining the events of the gait cycle and was used to generate the joint kinetics and kinematics used to validate the normal walking patterns. Data from foot markers were also required to detect foot-contact and foot-off events using auto-correlation. EMG, kinematic and force-plate data were synchronized and stored using VICON Nexus software in C3D format. Force-plate data were used to identify the time of initial contact and foot-off, augmented by auto-correlation for stepping off the force plate. All data were time-normalized to the gait cycle from one initial contact to the next of the same foot. They were exported to Noraxon MyoResearch-XP (master edition) (Noraxon Inc.,Scottsdale, Arizona, U.S.A.) for EMG post processing.

4.3.1 EMG data

a) Time normalisation. For each gait cycle, EMG and kinematic data were normalised to 100% of the gait cycle before export into Noraxon MyoResearch.

b) Removal of movement artefacts. Both surface and fine-wire EMG data were first band-pass 2nd order Butterworth filtered between 50 and1500 Hz. The band-pass filter values for surface EMG were higher than those stated in the SENIAM recommendations (which are for surface EMG only) because the EMG patterns detected by two types of sensors were going to be compared and so it was important to keep the same processing procedure. As far as we know, no definitive band-pass filter values for fine-wire EMG are available in the literature. Various low-pass and high-pass filter values were tried (Appendix 3). As the typical frequency range of the cable motion artefacts is between 1 and 50 Hz (Clancy, Morin, & Merletti, 2002). A high-pass cut of frequency of 50 Hz was used in this study to remove the low frequency artefacts with minimal interruption to the underlying EMG signal (Chapman et al, 2010; Semciw, Pizzari, Murley, & Green, 2013). This should not

significantly alter the correlation between surface and fine-wire EMG signals as the relationship between them was only altered significantly when the low cut-off frequency reaches 240 Hz (Brown, Brookham, & Dickerson, 2010).

c) Rectification and envelope detection. EMG data were then full-wave rectified and filtered with a 9 Hz low-pass filter as recommended by Shiavi et al. (1998), who found that this maintained at least 95% of signal power (Shiavi et al., 1998). This produced a linear envelope as recommended for EMG processing for dynamic contraction such as walking (Hermens & Merletti, 1999). Each completed gait cycle was exported to Excel (Microsoft Office 2013, Redmond, WA, USA).

d) Averaging. The data from six gait cycles which had consistent kinematic patterns and were within the normal range (Pinzone et al., 2014), were averaged to produce an individual ensemble average EMG profile.

4.3.2 Kinematics, kinetics and power

Kinematic and kinetic data and power were calculated using the Plugin Gait model in the Vicon software package (Nexus and Polygon). The PIG model is the most common model used to record the lower limb movements: kinematics, kinetics and power in clinical gait labs and using this will make the results most clinically applicable. However, this model regards the foot as a rigid segment, therefore the movement between hind-foot, mid-foot and forefoot are not reported. The use of multi-segment foot model may allow a better understanding of the muscle function. The signals were filtered by 4^{th} order Butterworth (zero lag) filter at 6 Hz cut-off frequency. This model is also known as Davis, Gage, Kadaba, Helen Hayes or Newington and also as the Conventional Gait Model. It has been developed and is mostly used by various researchers (Kadaba et al., 1989; Perry, 1992; Petersen, Hohmann, Stein, & Tillmann, 2002). The lower limbs are assumed to consist of seven

segments: pelvis, upper legs, lower legs and feet and three dimensional joint angles are defined from the relative orientation of adjacent segments. In the context of this thesis, these kinematic and kinetic data of the hip and knee joints are only used for reference purposes, to confirm normal patterns compared to a recent normal database (Pinzone et al., 2014). Only ankle kinetics and kinematics were focused on as EMG data were recorded from the muscles acting on the foot and ankle complex. Only one muscle acts on both the ankle and the knee.

Chapter 5 Normalisation of EMG during gait

This chapter aims to identify the most appropriate normalisation scheme for EMG processing in EMG measurements: fine-wire and surface sensors. Several available methods used in gait analysis will be described and compared. The experimental procedures were described in Chapter 4 and the experiment was carried out to select appropriate normalisation methods for use in future data collection based on between-subjects and between-session variability/ repeatability.

5.1 Background

5.1.1 What is normalisation?

Normalisation of data describes a range of data processing procedures designed to reduce known systematic variability between measurements. Successful normalisation allows the direct comparison of data between individuals/sessions without requiring any explicit consideration of the particular source of the variability. Time normalisation, for example, removes variability in stance time or gait cycle time, from gait graphs. The resampling of the curve using interpolation at a set number of intervals as a percentage (100 or 101) is the most widely used technique (Clayton & Schamhardt, 2001). In most gait analysis measurements the principal concern is to reduce the effects of the height or weight of the person under analysis (Hof, 1996). Regarding spatial temporal parameters, several normalisation approaches have been proposed such as non-dimensional numbers based on body size (Hof, 1996) and statistical techniques: offset, decorrelation and detrending (O'malley, 1996). Moreover, a net non-dimensional normalisation scheme has been developed to remove the dependence of the measures on height, weight and age to report an independent measure of gait efficiency (Pinzone, Schwartz, & Baker, 2016; Schwartz, Koop, Bourke, & Baker, 2006).

Normalisation of EMG data is different as the aim is to reduce the variability in the magnitude of the signals detected by sensors that arise from a variety of factors: the positions of sensors, muscles, individuals and sessions/days (Burden, 2010). Lehman and Mcgill (1999) demonstrated the misinterpretation of EMG measured from a different section of the same muscle within the same session when normalisation was not employed.

EMG normalisation is generally achieved by dividing the EMG from a reference contraction of the same muscle. The raw data are then reported as a proportion or percentage of the EMG reference value (Burden, 2010; Perry, 1992). The processing methods for these data from specific task and reference contractions are generally identical (Burden, 2010). This technique is suggested to be performed before comparison between sessions of the same individual or between subjects to avoid misinterpretation (Lehman & Mcgill, 1999).

In clinical gait analysis (CGA), normalisation is necessary for comparison within subjects over a period of time to detect changes and between subjects for diagnosis. The EMG from different healthy individuals should be normalised to establish a normative template (Yang & Winter, 1984). This template provides a normal pattern of EMG activity with expected normal variations for muscles, thereby facilitating the diagnosis of the pathological conditions. With large variations in the normative profile, mildly and moderately affected individuals may go undiagnosed. Various approaches have been used in previous studies with slightly different experimental settings. Therefore, it is necessary to investigate the effects of different normalisation procedures on variation in the normative EMG profile using surface and fine-wire sensors in order to determine the best method for clinical gait application where these sensors are used.

5.1.2 Evaluating the quality of normalisation

Measures of variability have been used as the criteria for selection normalisation techniques as the aim of normalisation is to improve repeatability (Knutson, Soderberg, Ballantyne, & Clarke, 1994; Yang & Winter, 1984). However, the indicator for repeatability has not yet been clearly identified. Therefore a range of parameters indicating variability have been widely used. The following parameters are used in gait analysis to describe the variability and repeatability of the data set.

Coefficient of Variance (CV) is the standard deviation divided by its mean value. It can be calculated between-subject (Leveau & Anderson, 1992; Limbird, Shiavi, Frazer, & Borra, 1988) or between-subject depending on the research purposes (Winter, 1984). It can be averaged across the gait cycle to give an overall indication of variability for any gait variables. It describes the spreading of data around the mean. However, CV is not a direct measure of absolute repeatability but the relative precision of measurement (Knutson et al., 1994). Between-sessions CV can be regarded as the measurement of repeatability because it is the estimation of pure measurement error within individuals and between-subjects CV accounts for the variation between individuals (Knutson et al., 1994). However, there may be a potential problem when the standard deviation (SD) is equal in two datasets with different average values. CV in data with a low average value can be higher. This could lead to misinterpretation.

Variance ratio (VR) can be calculated as the sum of the variance at each time point within a gait cycle divided by the total variance of the data. This is proposed by Hershler and Milner (1978). It is frequently used to assess repeatability (Burden, Trew, & Baltzopoulos, 2003; Kadaba, Wootten, Gainey, & Cochran, 1985). Richards, Thornton, and Delaney (2014) suggested the use of VR as an outcome measure to supplement the on-off EMG analysis of tibialis anterior and medial gatsrocnemius in order to identify the difference between

children with cerebral palsy and an age-matched control group. However, the details about the methodology employed in this study were unclear as it was reported in abstract form only. Burden et al. (2003) used CV and VR as between-session variability to compare the normalised and non-normalised profiles because the latter has been used widely in EMG repeatability studies to assess different sensors (Jacobson, Gabel, & Brand, 1995b; Kadaba et al., 1985) and between sessions (Kadaba et al., 1985). However, similar to CV, VR are a ratio to reference value and do not reflect the absolute nature of the signal (microvolt or percentage of normalised EMG). This may lead to misinterpretation as, when the actual value is small, the ratio is likely to be large. Moreover, without any information on the actual signal, the clinical interpretation of this ratio can be challenging.

Coefficient of multiple correlation (CMC) is a repeatability measure calculated from 'the positive square root of the adjusted coefficient of the multiple determination' (Kadaba et al, 1989). The adjusted coefficient of the multiple correlation is related to VR. It is calculated from one minus ratio of the variance between gait cycles and the total variance across the gait cycles. Therefore, if the total variance across the gait cycles is small, the CMC may be small and misinterpreted as affecting from poor reliability.

Intraclass correlation coefficient (ICC) is a variance ratio of true variance to total variance including the possible error components and facets of interests (Portney & Watkins, 2009). It is generally used to quantify the variance of a single measurement but can also be calculated at multiple time points over the gait cycle and averaged, as with the CV. It indicates the similarity between trials compared with the differences between subjects (Francis, 1986). Wills, Hoffer, and Perry (1988) demonstrated six forms of ICC for a reliability study with a selection guideline. This technique additionally informs an agreement of the mean when compared with Pearson's correlation coefficients which are used in a few reliability studies on EMG (Di Fabio, 1987; Giroux & Lamontagne, 1990; Horstmann,

Gollhofer, & Dietz, 1988; Knutson et al., 1994). Similar to the aforementioned measures, this measure does not provide the information about the actual signal to allow direct interpretation in practice.

In the studies where different normalisations were compared, Knutson et al. (1994) suggested the use of VR and ICC as the statistical indicators for repeatability within day performance as CV showed only the variability between subjects within a group and not the reliability (although there is no particular reason why CV should not be calculated based on a series of repeat measurements made on an individual). ICC can also inform both association and agreement as the systematic changes in measurement would also affect it. However, there is no true gold standard, for the measures of reliability or criteria suggest the clinical meaning of these values.

The use of *Standard Error of the Measurement (SEM)* is proposed as it has a clinical meaning, allowing a clinical interpretation. It is the variation between measurements of the same quantity on the same individual (Bland & Altman, 1996). It is one of the common statistical methods employed to express response stability; estimating the standard error in a set of repeated scores and establishing reliability (Portney & Watkins, 2009). When this value is calculated from a group of subjects with regard to test-retest repeatability, it can provide an estimation of the range of measurements that can be expected on retesting based on a confidence interval (CI) as following:

95% CI = Observed value +/- 1.96 (SEM) (Portney & Watkins, 2009)

SEM is one of the commonly used methods in repeatability studies to quantify the errors in the original measuring unit such as degree or microvolt (Stratford & Goldsmith, 1997). It was the primary measure used by Mcginley, Baker, Wolfe, and Morris (2009) for the most comprehensive and widely cited systematic review of variability in walking.

The SEM comprised several sources of variability; for example, in gait analysis, the main sources of variability are random and systematic differences in marker placement by different analysts and the differences between a participant's walking trials. These are regarded as variance components (Baker, 2013). Schwartz et al. calculated SEM to present sources of variance within session, between sessions and between analysts for kinematic data (Schwartz, Trost, & Wervey, 2004). This information is useful for quantifying the quality of marker placement by analysts, leading to improved techniques. Using SEM to identify the sources of variance will hopefully lead to improved data measurements later. For EMG, it has been used to assess reliability in neck and trunk muscles, e.g. Burnett, Green, Netto, and Rodrigues (2007); Dankaerts, O'sullivan, Burnett, Straker, and Danneels (2004); Netto and Burnett (2006), more extensively than lower limb muscles (Norcross, Troy Blackburn, & Goerger, 2010). For ankle muscle such as the tibialis posterior and the tibialis anterior, only one study has employed this parameter but it is limited to EMG in a stance phase in a patient sample (rheumatoid arthritis) (Barn et al., 2012).

In summary, whilst a range of relative measures of repeatability (CV, VR, ICC, CMC) have been reported in the past, these all give ratio values that are difficult to interpret in relation to clinical measurements. Absolute measures, being reported in the units of measurement, give a far more direct and clinically useful indication of variability. The SEM is the only absolute measure of repeatability that has been commonly used in gait analysis and will thus form the focus of this study. A range of relative measures will also be calculated to allow comparison with the finding of the earlier literature.

5.1.3 EMG normalisation methods in the literature

There are many recently published works that evaluate the different normalisation methods of EMG amplitude but these studies and reviews provide inadequate clinical justification for

the selection of the appropriate technique for CGA (Burden, 2010; De Luca, 1997; Kamen & Gabriel, 2010; Knutson et al., 1994; Perry, 1992; Robertson, 2004; Yang & Winter, 1984). Regarding amplitude normalisation, there were eight normalisation methods reported in the recent literature, classified by their denominators for comparisons of EMGs over the last 25 years, described by Burden (2010). For gait analysis, the normalisation can be classified into *self-normalisation* which relates to a characteristic within the signal being recorded such as dynamic peak amplitude or dynamic mean amplitude during gait cycles and *external normalisation* which relates to a characteristic of a separate test such as maximum voluntary isometric contraction (MVIC).

Denominators which have been used for self-normalisation include *Mean* which is the average EMG from the gait (Winter & Yack, 1987; Yang & Winter, 1984), and *Peak* which is the maximal EMG from the specific task - walking (Jacobson et al., 1995b). These are usually derived from ensemble averaged EMG (Burden et al., 2003).

For external normalisation, *Maximum voluntary contraction (MVC)* is the most popular in gait studies (Dubo et al., 1976; Fuglevand et al., 1992; Hermens & Merletti, 1999; International Society of Electrophysiological Kinesiology. Ad Hoc Committee, 1980). The primary assumption for MVC is that all motor units are firing at their highest rate during a maximum contraction (Bigland & Lippold, 1954; Merton, 1954). The EMG reference recorded from MVC in healthy adults theoretically reveals the proportion of the maximal activation capacity of the task that the observed EMG represents (Allison, Marshall, & Singer, 1993; Yang & Winter, 1984), so the EMG during a specific task is reported as a proportion of maximal capacity. There are several options for collecting the MVC EMG: isometric or dynamic (Burden, 2010), manual muscle test (Cavanagh & Komi, 1979; Murley, Buldt, et al., 2009) or subject instruction to lie the joint in a set position (Konrad, 2006). The Journal of Electromyography and Kinesiology‘s guidelines for reporting

research and the Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM) project accept the use of the MVC reference value, provided that the factors influencing EMG have been reported; for example, the joint angle and/or muscle length during MVC, and the rate of rise of force implying the use of MVC from both isometric and dynamic conditions.

Other external normalisations include:

- *Submaximal isometric voluntary contraction (Isometric-$_{subMVC}$)* -the highest EMG obtained from an isometric voluntary contraction which generates a steady force that is well below the maximal force (Burden, 2010).
- *Submaximal dynamic voluntary contraction (Dynamic-$_{subMVC}$)* - the highest EMG obtained from a non-isometric contraction which generates a constant force that is well below the maximal force (Burden, 2010).
- *Arbitrary angle isometric maximal voluntary contraction (Isometric $-_{arbMVC}$)* - the highest EMG obtained from an arbitrary mid-range joint angle (Burden, 2010).
- *Angle specific maximal isometric voluntary contraction (Isometric-$_{specMVC}$)* - the highest EMG obtained from a maximal isometric voluntary contraction with the same muscle action, and a joint angle or muscle length as the task EMG (Burden, 2010).
- *Angle specific maximal dynamic voluntary contraction (Dynamic-$_{specMVC}$)* - the highest EMG obtained from a non-isometric voluntary contraction with the same muscle action, and joint angle or muscle length as the task EMG (Burden, 2010).
- *Angle and velocity specific maximal isokinetic voluntary (Isokinetic-$_{specMVC}$)*- the highest EMG obtained from a maximal isokinetic voluntary contraction

with the same muscle action, joint angle or muscle length, and angular velocity or rate of change of length as the task EMG (Burden, 2010).

- *Maximal M-wave amplitude* can also be used for normalisation. This technique involves the stimulation of a peripheral motor nerve to activate most /all of the motor neurons. The result is a maximal EMG capability- a synchronous signal M-wave. The negative-peak–to positive-peak amplitude or area will be calculated to normalise the EMG data (Robertson et al., 2013). However, this method is not commonly applied in gait studies.

These terms are not always explicitly or consistently used in the literature and the details of how normalisation was achieved in the studies are frequently neglected.

5.1.4 Selecting the appropriate normalisation

Normalisation is an important technique for processing EMG as it allows comparisons between subjects and sessions for diagnosis and changes as follow-up appointments and evaluation of the treatment. The best normalisation should result in high reliability and low variability between homogenous subjects and sessions. The normalisation should not increase the natural variability between individuals but minimise the systematic or measurement variability to allow the detection of pathological features. The settings should be feasible for pathological groups. The systematic review in Chapter 3 revealed that 19 studies employed normalisation techniques. The denominators included *maximal voluntary isometric contraction (MVIC)* (frequently, the joint angles were not described, so it was difficult to decide whether it was Isometric $-_{arbMVC}$ or Isometric-$_{specMVC}$) (10 studies), mean amplitude during an isometric contraction (1 study), mean amplitude during a gait cycle (2 studies) and peak amplitude (6 studies) during a gait cycle, but none of them provided an

adequate justification for their selected normalisations. MVIC has great potential for clinical use as the normalised EMG would be presented as a proportion of the maximal muscle activation capacity, while the mean and peak would be presented as a proportion of the EMG relative to the reference values.

Many studies compared different normalisations for superficial muscles. Burden et al. (2003) found small differences in amplitude and patterns between Isometric $-_{arbMVC}$ and Isokinetic-$_{specMVC}$ but the between gait cycle variability of the Isokinetic-$_{specMVC}$ was higher than that of the Isometric-$_{arbMVC}$. The Isokinetic$_{specMVC}$ required more time and effort to produce and part of the gait cycle may not be normalised for some individuals whose range of muscle length during gait exceeded the recorded range during calibrated contractions: isometric, concentric and eccentric contractions, as seen in Burden's work. There was also a comparison between Isometric $-_{arbMVC}$ and segment weight dynamic movement or what Burden terms as Dynamic$_{-subMVC}$ (Nishijima, Kato, Yoshizawa, Miyashita, & Iida, 2010). The shapes of the EMG profiles were similar and there was no significant difference in CV or VR between the two techniques. The segment weight dynamic movement was relatively complicated and time consuming. Therefore Isokinetic$_{specMVC}$ and segment weight dynamic movement may not be more effective or suitable than Isometric $-_{arbMVC}$ or MVIC for gait researchers.

There are several techniques available for obtaining MVIC. Murley, Menz, Landorf, and Bird (2010) compared two normalisations: MVIC applied by manual muscle test (MMT) and peak EMG at the fastest speed of walking on lower limb muscles using both surface and fine-wire sensors. Overall, the peak normalisation resulted in lower between-subject variability within session and between-session variability than MVIC normalisation in most studied muscles. The MVIC normalised EMG variability was lower than that of the non-normalised EMG within session for surface EMG but, for fine-wire EMG, the variability of

MVIC normalised EMG between subjects was similar to that of the non-normalised EMG. Yang and Winter (1983) also compared electromyography repeatability between the Isometric $-_{arbMVC}$ and Isometric-$_{subMVC}$ and suggested that the latter was more repeatble based on CV and ICC which was used as an indicator of repeatability. The external normalisations can be carried out by MMT or machines such as quantitative muscle testing (QMT) using a dynamometer. In regard to the dorsiflexors, QMT was found to be more reliable and easier to implement than MMT (Escolar et al., 2001). Moreover, the dynamometer allows identical positions of the participants between sessions as well, which may increase the repeatability of the normalised EMG signal.

The effect of choosing MVIC, peak and mean on the repeatability of EMG from lower limb muscles is still inconclusive. For knee extensors and flexors, the MVIC reduced the between-subject variability of the raw EMG but the variability was still higher than the peak and mean normalisations (Burden et al., 2003). On the other hand, it was found that the MVIC resulted in the greatest repeatability within day performance compared to a mean and a peak in gastrocnemius during walking (Knutson et al., 1994). Murley et al (2010) found that the peak normalisation obtained from walking at the fastest resulted in lower between-subject variability within session and between-session variability than MVIC normalisation (MMT) in most studied muscles. This may suggest that the effect of MVIC is specific to muscles.

It has been found that a mean and a peak EMG for normalisation effectively reduced between-subject variability compared with Isometric$_{subMVC}$ and Isokinetic $_{specMVC}$ (Burden et al., 2003; Yang & Winter, 1984), whereas Isometric$_{SubMVC}$ and Isokinetic$_{SpecMVC}$ tended to increase between-subject variability. Shiavi, Bugle, and Limbird (1987a) showed the similarity between the EMG profiles, normalised by peak and mean. However, the standard deviation (SD) in the peak normalised profile was more uniform than that by mean during

the active periods. The SD in the mean normalised profile was less than that normalised by the peak when the muscles were inactive. Burden et al. (2003) concluded that the most homogenous ensemble averaged profiles may result from these techniques but the true variations within a group were removed. This may lead to a false positive clinical diagnosis, as Knutson et al. (1994) caution. Also, when normalised to the mean/peak of the gait cycle, the amplitude does not imply the required level of muscular activity during gait but the relative activity to the mean or peak in different phases.

In addition, the repeatability of the EMG detected by different sensors is important information for CGA. This may be calculated using the measures of variability between different healthy subjects and/or different sessions. A few studies have assessed the repeatbility of fine-wire and surface EMG sensors on superficial lower limb muscles using different normalisations within and between sessions (Bogey, Cerny, & Mohammed, 2003; Cavanagh & Komi, 1979; Jacobson et al., 1995b; Kadaba et al., 1985) but the repeatability of fine-wire and surface sensors are still controversial due to the different measures used preventing direct comparison, different fine-wire application techniques including use of cross-talk tests and different time-periods between sessions. Furthermore, the reliability of EMG may be specific to both muscles and sensor types due to the anatomical position of muscles which may be subjected to crosstalk from the surrounding muscles and the different detecting sites of the sensors.

Whilst this section has so far focussed on the potential of normalisation to reduce variability, the normalisation method may also make the data more clinically meaningful. Normalisation with respect to some form of maximum voluntary contraction may be regarded as an indication of what proportion of muscle capacity is being used at any one time in the gait cycle. Normalisation with respect to a known sub-maximal voluntary contraction, by contrast, may give an indication of the proportion of that force being exerted. The

complexity of muscle physiology in general and the relationship between the EMG signal and force production (see for example Chapter 2 of Lieber (2010)) however, suggests that caution should be excercised when making simplistic interpretations on the basis of how data are normalised. Normalisation to the peak or mean signal over the gait cycle will result in all signals being of a similar magnitude but this may be particularly useful if the pattern of activity (rather than its magnitude) is the main focus of clinical investigation. The relative merits of these other aspects of normalisation should be taken into consideration alongside the different effects on the variability of the measurement.

In summary, a wide range of methods have been suggested for the normalisation of EMG signals for a range of different applications. Although a wide range of studies has been conducted, none has given definitive guidance regarding the best method for clinical gait analysis. Of the methods related to standardised muscle contractions, the most practical is MVIC, which also potentially provides the most meaningful clinical interpretation and this will be the focus of the investigations in this chapter. Peak and mean normalisations will also be investigated as they showed promising results with regard to improved repeatability and reduced variability in previous reports.

5.1.5 Research question

i) In healthy participants, what effect does the normalisation method have on the between-subjects and between-sessions repeatability of linear envelope EMG signals collected with fine-wire or surface sensors?

ii) In light of this and other differences between the normalisation schemes, which is the most appropriate for future clinical use?

5.2. Method

The recruitment of participants, procedures and data processing were described in Chapter 4. This chapter described the normalisation of EMG data and calculation of measures for different normalisation factors: MVIC, mean and peak for each participant during each session at self-selected speeds. The MVICs were carried out on an isometric dynamometer set consistently during all sessions. The EMGs were recorded for three seconds with verbal encouragement and a visual force display on the screen. The mean and peak normalisation factors were calculated from the included gait cycles and an ensemble average, for each sensor, for each individual and for each session. Then the between-session variability (indicated by SEM) was calculated based on the individual ensemble average normalised EMGs from two sessions and the between-subject variability of the grand ensemble average normalised EMGs in session 1 was calculated for each normalisation technique.

5.2.1 Normalisation: Calculation of reference amplitudes

a) Maximum voluntary isometric contraction (MVIC)

MVICs were carried out while the participant was stabilised using shoulder and thigh straps, on the Kin Com dynamometer (Chattanooga Group, Hixson, TN) chair. The lateral malleolus was carefully aligned with the axis of rotation of the dynamometer and the foot was secured to the lever arm of the dynamometer using a padded velcro strap (Figure 5.1-2). The participants were asked to perform three maximum contractions at the ankle (dorsiflexion, plantar flexion and inversion) for three seconds. They were instructed to push as hard as they could in each position against a plate on the dynamometer with verbal encouragement and feedback from the force level visualised on the computer screen next to them. There was a minute rest between each contraction to avoid fatigue. The dynamometer was used to standardise the muscle length for isometric contraction, minimise the difference

between joint positions between sessions and maintain a constant applied resistance on the participants.

i) Each recorded EMG was processed identically as described in 4.3.1 (a-b).

ii) The average EMG signal was calculated separately from the middle one second of each of the three MVCs.

iii) The largest of the three values was chosen as the MVIC normalisation factor for each participant for each session.

Figure 5.1 Position for inversion

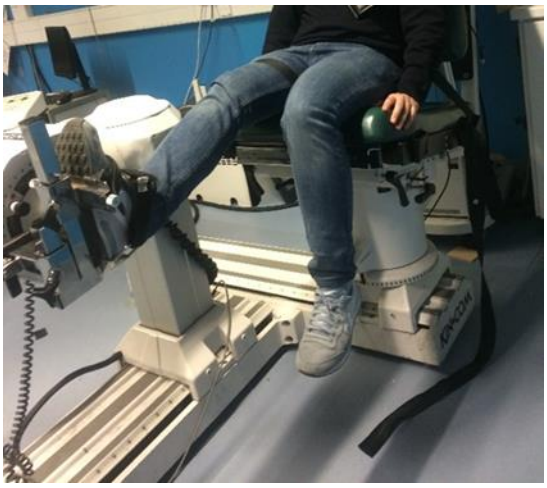


Figure 5.2 Position for dorsiflexion and plantarflexion

b) The mean and peak values

The mean EMG value was an average from all 1% intervals of the six gait cycles included for an ensemble average EMG profile (4.3.1.d). The peak EMG value was calculated from an average of the maximum EMG values from 1% intervals of the six included gait cycles. These two values were calculated for each sensor, each participant, and each session at the self-selected speeds.

5.2.2 Walking EMG data

a) Surface EMG from the tibialis anterior, medial gastrocnemius and proximal fine-wire EMG from the tibialis anterior, medial gastrocnemius and the tibialis posterior were processed as described in Chapter 4-4.3.1(d). The proximal fine-wire data were chosen as described in Chapter 6, and there was no observable difference between two sites of fine-wire placement in the same experimental setting.

b) In Excel (Microsoft Office 2013, Redmond, WA, USA), time normalised EMG (4.3.1 (d)) were divided by their corresponding three normalisation factors: MVIC, peak and mean for each participant, each sensor and each session.

c) Then the ensemble average for each sensor and each session for all participants was calculated for non- normalisation data and each normalisation scheme, for each sensor and each session to assess inter-subject variability.

d) To visualise and compare normalised EMG profiles, the grand ensemble average signals (5.2.2(c)) were scaled to their corresponding mean value for non-normalisation and each normalisation, each sensor and each session (Figure 5.3).

e) The standard deviation (SD) of the grand ensemble averages (c) for each normalisation and non-normalisation were scaled by their mean values for each point in the gait cycle for each sensor and each session (Figure 5.4).

f) A single representative standard deviations was calculated by averaging (e) across the gait cycle for each normalisation, each sensor and each session (Figure 5.5).

g) CV, VR, CMC were calculated from the data (c) for each normalisation, sensor and each session to assess between-subject variability.

h) CV, CMC and SEM were calculated from the data (b) for each normalisation, each participant, and each sensors between two sessions to assess between-session variability

i) The SEM for each muscle were scaled by the mean values from (b) to allow a direct comparison between different normalisation schemes.

5.2.3 Assessment of variability

a) Within session: between-subject variability

All normalisations applied in this experiment used a single number as the denominator, so the between-subject variability of normalized EMG signals in the same session was identical to non-normalised signals. The between-subject variability within the same session can be affected by the normalisations: peak, mean and MVIC, so SD was calculated as an index of measurement variability of the normalised and non-normalised EMG profiles in the first session from 5.2.1.6. CV, VR and CMC were also calculated for comparison with the previous literature.

b) Within subject: between-session variability

SEM was also calculated for the participants between sessions to determine the most effective normalisation which minimised the within-subject, between-session variability. SEM was estimated from the SD and ICC (2,1) which was used as the repeatability of measurement from the data 5.2.1.2 . In this model, the participants and measuring sessions were considered to be randomly chosen. The analysis of ICC for this paper was generated using Excel (Microsoft office 2013, Redmond, WA, USA), and the Real Statistics Resource

Pack software (Release 3.5, Copyright 2013 – 2015, Charles Zaiontz) for every 1% interval of the gait cycle. The CV, VR and CMC were also calculated over 12 gait cycles for each individual (six gait cycles from each session).

5.3 Result

In this study, 11 participants were recruited but only 10 (age 34 ±4 years old, 4 females and 6 males, height 1.67 ± 0.10m, and weight 70 ±13 kg) completed two testing sessions. One of the participants could not attend due to lack of availability. Therefore, data from 10 participants were used for between-session variability whereas the between-subject variability was calculated from 11 participants. The average speeds for Session 1 and Session 2 (at least two weeks after the first session) were similar: 1.2 ±0.2 m/s and 1.1±0.2 m/s respectively.

5.3.1 Effects of normalisation on within session: between-subject variability

Figure 5.3 shows the EMG profiles of the tibialis anterior muscle with ±1SD from a scaled grand ensemble average from 11 participants at self- selected speed as an example of how the standard deviation bars are affected by different normalisation methods. Without scaling, MVIC results in a minimal mean amplitude profile among the three techniques. Scaling normalised EMG by theirs means allows a direct comparison between normalisation methods. The clinical use of these data will thus be to compare the patterns rather than the magnitude.

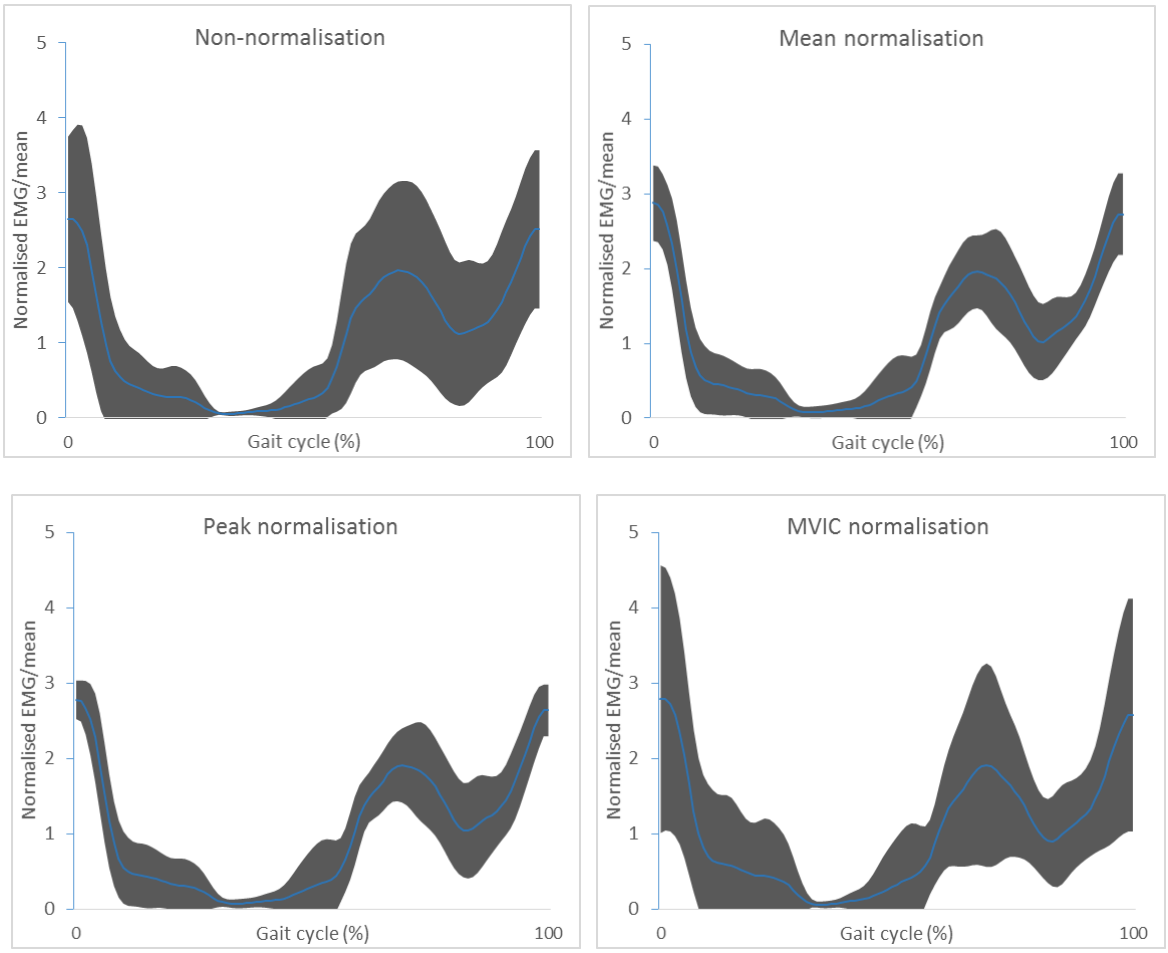


Figure 5.3 Grand ensemble averages of fine-wire EMG of tibialis anterior (TA) from session 1 with different normalisations scaling to their means (the blue line is the average from 11 participants and the grey areas are ± SD)

The SD/mean profile shows greater variation during the active period of the EMG profile for all normalisation techniques (Figure 5.4 a). The ratios of SD to the mean calculated from the peak and mean normalisations are lower than those from MVIC normalization and non-normalisation (Figure 5.4 a and b). Similar behaviour was observed for all muscles and sensors and therefore only the bar charts will be considered for the other muscles.

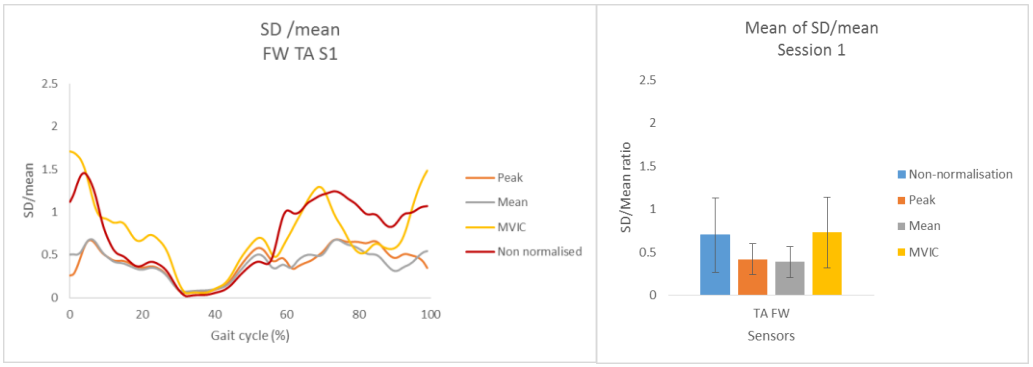


Figure 5.4 (a) SD/Mean across the gait cycle and (b) Mean of SD/mean from fine-wire (FW) EMG of tibialis anterior (TA) session 1

Mean and peak normalisations reduce variability across all muscles and sensors (Figure 5.5). Mean normalisation shows a slightly better performance than peak normalisation and results in a reduction from non-normalised values of between 18% in the tibialis anterior (surface) and 62% in the tibialis posterior (fine-wire). By contrast, the MVIC normalization increases the variability more than mean or peak normalisation in all muscles and, in the tibialis anterior (fine-wire and surface) and medial gastrocnemius (surface), actually results in an increase in variability with respect to non-normalised data. Between sensors, MVIC increases the variability in surface EMG more than fine-wire EMG.

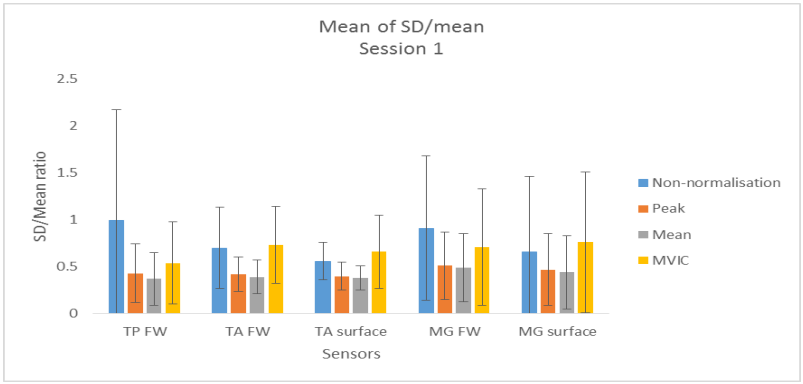


Figure 5.5 Mean of SD/mean in session 1 (FW is fine-wire EMG, tibialis posterior (TP), tibiaiis anterior (TA) and medial gastrocnemius (MG))

CV, VR and CMC show essentially the same trends as the SD/mean with subtle differences reflecting the nature of the different measures. Mean and peak normalisations reduce CV and VR and increase CMC compared with the non-normalisation and MVIC technique. Mean normalisation results in a reduction of CV and VR from non-normalised values of

between 23% of CV in the tibialis anterior (surface) and 85% of CV in the tibialis posterior (fine-wire) (Figure 5.6); 11% of VR in the tibialis posterior (fine-wire) and 35% of VR in the medial gastrocnemius (fine-wire) (Figure 5.7). Mean normalisation increases CMC of the non-normalisation between 6% in the tibialis posterior (fine-wire) and 22% in the medial gastrocnemius (fine-wire) (Figure 5.8).

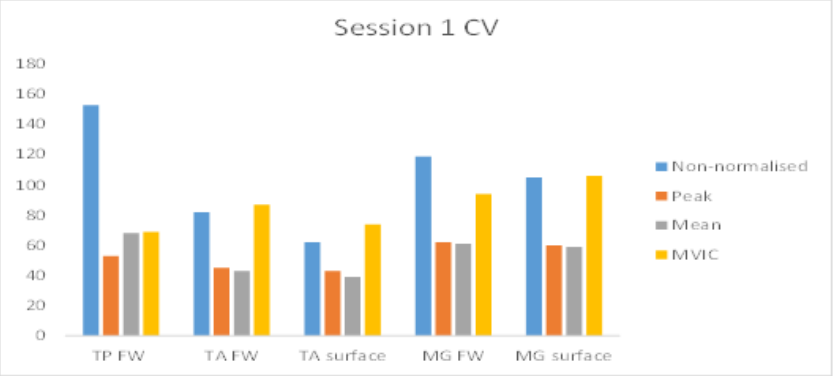


Figure 5.6 Between-subject CV in session 1 (FW is fine-wire EMG, tibialis posterior (TP), tibialis anterior (TA) and medial gastrocnemius (MG))

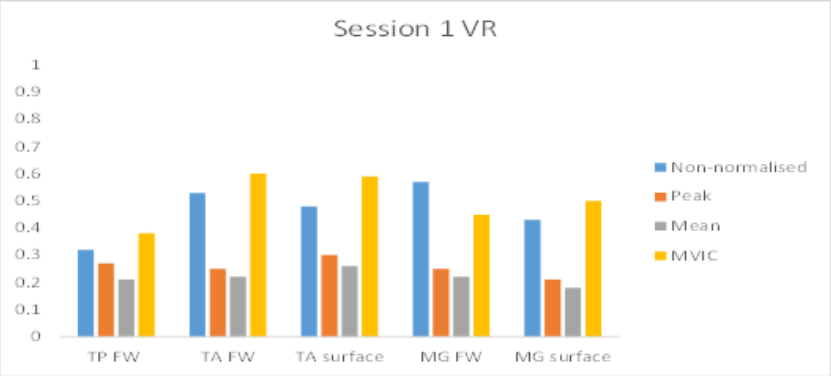


Figure 5.7 Between-subject VRc in session 1 (FW is fine-wire EMG, tibialis posterior (TP), tibialis anterior (TA) and medial gastrocnemius (MG))

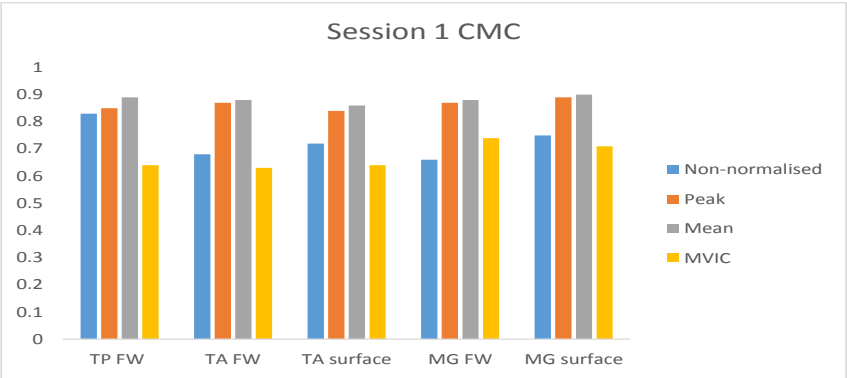


Figure 5.8 Between-subject CMC in session 1 (FW is fine-wire EMG, tibialis posterior (TP), tibialis anterior (TA) and medial gastrocnemius (MG))

5.3.2 Effects of normalisations on within subject: between-session variability

SEM are presented as proportional to their corresponding mean values during walking, (Figure 5.9) which results in a reduction of variability from non-normalised up to 42% of the mean value for the tibialis posterior (fine-wire) and shows a slightly better performance in fine-wire sensors than peak normalisation (difference <5%). The MVIC normalisation increases the variability more than mean or peak normalisation in all muscles and, in surface EMG of the tibialis anterior and medial gastrocnemius actually results in an increase in variability with respect to non-normalised data (Figure 5.9).

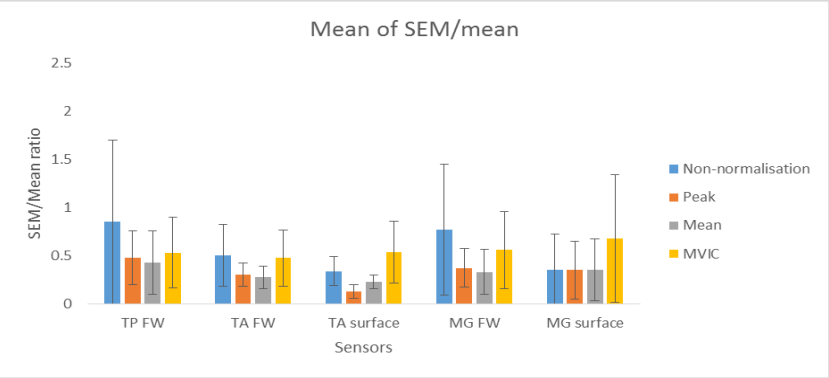


Figure 5.9 Mean of SEM/mean across the gait cycle from different normalisations (FW is fine-wire EMG, tibialis posterior (TP), tibialis anterior (TA) and medial gastrocnemius (MG))

CV, VR and CMC for between-session variability show the same trends as the SEM/mean. Peak and mean normalisations reduce CV and VR and increase CMC compared with non-normalisation and MVIC normalisation. Mean normalisation results in a reduction of CV and VR from non-normalised values of between 5% of CV in the tibialis anterior (surface) and 24% of CV in the medial gastrocnemius (fine-wire) (Figure 5.10); 3% of VR in the medial gastrocnemius (surface) and 13% of VR in the medial gastrocnemius (fine-wire) (Figure 5.11). Mean normalisation increases CMC of the the non-normalisation between 2% in all surface EMG and 7% in medial gastrocnemius (fine-wire) and tibialis posterior (fine-wire) (Figure 5.11).Similar to inter-subject variability, CV and VR seem to be sensitive to normalisation whereas CMC tends to hide the differences. Moreover, differences between

the surface and fine-wire sensors seem less clear regarding the relative measures of between-session variability. The difference between the surface and fine-wire non-normalised EMG are small in CMC (3-6%), VR (5-11%), CV (16%) whereas SEM shows a 16-42% difference.

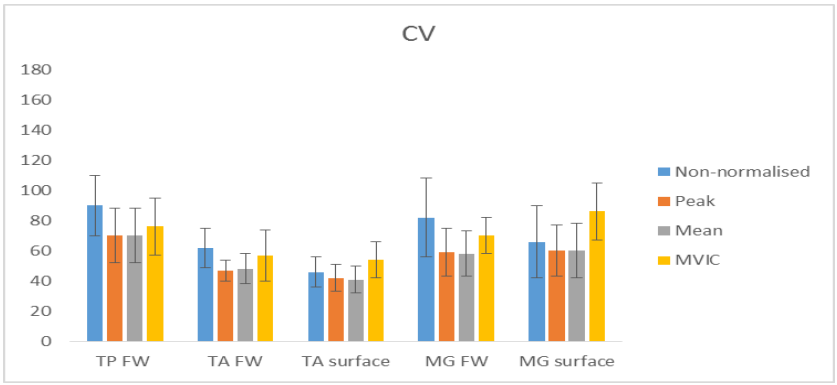


Figure 5.10 Between-session CV (FW is fine-wire EMG, tibialis posterior (TP), tibialis anterior (TA) and medial gastrocnemius (MG))

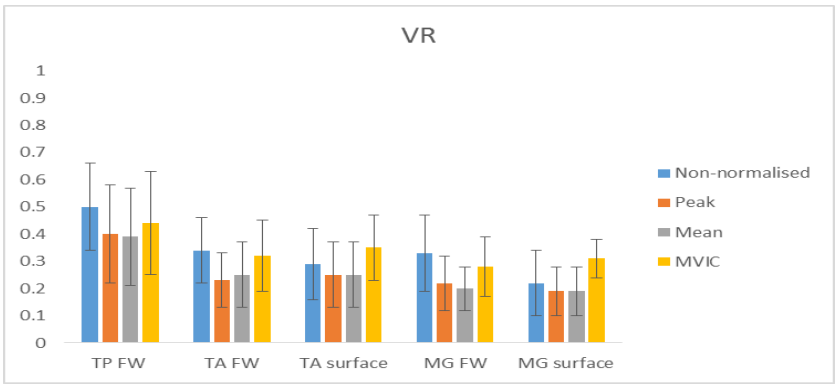


Figure 5.11 Between-session VR (FW is fine-wire EMG, tibialis posterior (TP), tibialis anterior (TA) and medial gastrocnemius (MG))

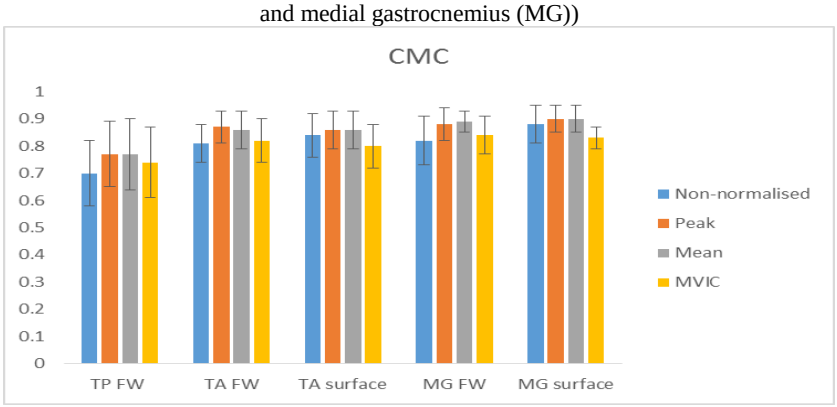


Figure 5.12 Between-session CMC (FW is fine-wire EMG, tibialis posterior (TP), tibialis anterior (TA) and medial gastrocnemius (MG))

5.4. Discussion

5.4.1 Within session: between-subject variability

a) Different normalisation procedures

The large variability represented by high ratio of SD/mean, large CV and VR and low levels of similarity between waveforms indicated by low CMC are observed on non-normalised grand ensemble average EMG profiles (Figure 5.5-8). The average SD/mean in non-normalised fine-wire EMG of tibialis posterior and medial gastrocnemius are approximately the same as the mean signal. The large between-subject variability of surface EMG were previously reports: 40-50% CV in the tibialis anterior and 55-113% in the medial gastrocnemius (Murley et al., 2010; Winter & Yack, 1987; Yang & Winter, 1984). With the large variation in the normative template, the detection of clinical difference in EMG profile may be difficult because they may be considered as normal variation (Murley, Buldt, et al., 2009). Therefore the normalisation of the EMG signals is necessary for comparison between individuals to detect clinically important differences. Although it is difficult to compare measurements of different kinds made in different units, it would appear that EMG data are inherently more variable (between subjects and sessions) than is the case with kinematic data (Mcginley, Wolfe, Morris, Pandy, & Baker; Mcginley et al., 2009). Pinzone et al. (2016) recently commented that variability amongst kinetic parameters (even after non-dimensional normalisation) can also be appreciable and this might suggest that there is greater variability in both EMG and kinetic measurements than in the kinematics to which they contribute. Given that being able to distinguish between normal and abnormal is one of the key requirements of any biomechanical measure (Baker, 2006), the potential for normalisation schemes to reduce variability is probably more important than other aspects such as making the measurements more clinically meaningful.

SD and SEM are common to determine *minimal detectable difference (MDD)* which is 'the smallest amount of change that can be considered above the threshold of error expected in the measurement'(Portney & Watkins, 2009) or variation expected between healthy participants or between sessions. They are various across the gait cycle: the greater value is reported during active period and the smaller value is reported during inactive period in healthy adults. As $MDD = z * SEM * \sqrt{2}$ (at 95% confidence interval z = 1.96) (Portney & Watkins, 2009), the inactive period of the EMG profile is the most sensitive to detect presence of additional activity related to pathological changes. However, MDD is not the same as *minimal clinically important difference (MCID)* which is 'the smallest changes in an outcome measure that is perceived as beneficial by the patients, and that would lead to a change in patient's medical management, assuming an absence of excessive side effects and cost'(Portney & Watkins, 2009).

Mean and peak normalisation reduce between-subject variability across all muscles and sensors, as indicated by the average SD/mean, CV, VR and CMC. Mean normalisation shows a slightly better performance than peak normalisation (small margin 3-5%) and resulting in up to a 62% reduction from the non-normalised values in the tibialis posterior (fine-wire), and less reduction (approximately 18%) in the tibialis anterior (surface). The CV reduction of EMG variability on the tibialis anterior during gait using peak and mean values was also reported by Yang and Winter (1984) and Winter and Yack (1987). MG. Burden et al. (2003) also agreed on the greater effectiveness of peak and mean normalisations compared to MVIC on more proximal muscles during gait. In agreement with Shiavi et al. (1987a) and Burden et al. (2003), the VR of EMG normalised by mean was lower than that that of the peak and CMC of the EMG profiles normalised by mean was higher than the MVIC. The results suggest that the mean normalised EMG ensemble average was the most homogenous among the other normalisations but only marginally better than

the peak normalisation. However, there is a drawback associated with using the mean normalisation method since some participants may have spikes of EMG amplitude in some gait cycles, resulting in relatively high peak values (Burden et al., 2003). The mean value would conceal this effect (Burden et al., 2003).

MVIC normalisation is the least effective method for reducing between-subject variability, occasionally resulting in greater between-subject variability than non-normalised EMG (Figure 5.5). Our MVIC setting is different from previous studies in an attempt to obtain the maximum activity of the muscles while the lower limb joints are stabilized by a dynamometer but the results were similar to previous reports (Dankaerts et al., 2004; Murley et al., 2010). MVIC may be considered another source of errors (Yang & Winter, 1984). This increase in variation may be caused by different levels of muscles in proportion to their MVIC required to walk between individuals as they may have different fitness levels. Additionally, at a high level of contraction, there may be muscle substitution to maintain the generated force. For example, the medial/lateral gastrocnemius and soleus may be involved, resulting in changes to the EMG even though force is constant.

b) Different sensors

In non-normalised, peak normalised and mean normalised EMG profiles, the surface EMG profiles tended to show slightly less variation than the fine-wire EMG profiles, as indicated by the lower ratio of SD/mean, CV, and VR and the higher CMC (Figure 5.5-5.8). There may be different numbers of motor units in the relatively small detecting volume of fine-wire during walking as the required force was changed. However, the difference between sensor types is very small and may be caused by experimental errors. The similar responses of surface and fine-wire EMG to the mean and peak normalisations confirms that the fine-wire EMG from deep muscles can be comparable to other surface EMG.

On the other hand, when normalised by MVIC values, the variability increased in the tibialis anterior and medial gastrocnemius, particularly in surface EMG. Figure 5.5 shows that the ratio of SD/mean after MVIC normalisation is higher than raw EMG. Bogey et al. (2000) also showed the longer active period of normalized surface EMG of the soleus when compared with the MVIC normalized fine-wire EMG. They suggested that MVIC normalised EMG from a fine-wire sensor was accurate and provided more precise temporal data compared with surface sensors. A possible assumption could be that the surface sensors may detect signals from the adjacent muscles due to their relatively large detection volume compared with fine-wire sensors. Additionally, the different activity patterns between healthy individuals during walking in those muscles may account for additional variations. To our knowledge, no study has investigated the effect of different normalisations on different types of EMG sensors during gait on the tibialis anterior and medial gastrocnemius. The result of this study may aid the selection of appropriate normalisation based on the use of sensor types specific to certain muscles.

5.4.2 Within subject: between-session variability

a) Different normalisations

The average speed of 1.14 m/s during the second testing session was within the range of 0.95-1.43 m/s which was the speed range measured from the first testing session. None of the participants show apparent gait deviations during walking in either session. The recent review showed the high reliability of the three-dimensional kinematic gait measurement (less than 5º in healthy subjects)(Mcginley et al., 2009). Therefore the EMGs between sessions should not be affected by the different speeds or different walking patterns but should be affected by the natural variability of EMGs between days due to the reapplication of electrodes and individual conditions.

The control of between-session variability (systematic measurement) would allow a comparison within subject over a period of time. This affects the clinical evaluation of the treatment protocols. Between two sessions, the variability of the tibialis anterior and medial gastrocnemius without any normalisations was similar to that in the previous study. Kadaba et al. (1989) reported the repeatability of non-normalised EMG between sessions; CV: 50%±6% for surface EMG of TA and 58% ±9 % for surface EMG of MG,CMC: 0.832 ±0.06 for TA and CMC of 0.875±0.04 for MG. These between-sessions should be reduced to allow the clinical application.

In our study, mean normalisation effectively reduce the between-session variability from 11% in the tibialis anterior (surface) and 44% in the medial gastrocnemius (fine-wire). Peak normalization is equally good with small margin differences (up to 5% difference) and in some cases (fine-wire sensors) peak normalisation is more effective. The effects of the mean and peak are similar to inter-subject variability.

Between two sessions, the SEM/mean for the tibialis anterior and the medial gastrocnemius are considerably higher for the MVIC than for the other techniques (Figure 5.9). During the active phase, the SEM/mean for MVIC is often greater than that for non-normalized EMG, possibly because the recorded EMG during MVIC testing was not the true MVIC despite the verbal encouragement and training. Also, muscle substitution and co-contraction may be considered as additional varying factors.

b) Different sensors

Regarding the normalisation of fine-wire EMG, the mean denominators result in higher repeatability (lower SEM/mean) but the peak denominators result in a slightly lower ratio in surface EMG (Figure 5.9). Similar to inter-subject variability, the MVIC normalization tends to increase the inter-session variability. The surface EMG appears to be highly affected by MVIC normalisation as their CVs are higher than that of non-normalised signal, possibly

because of the larger detection volume of the surface sensors collecting the signals from the surrounding muscles when there may be different muscle substitution and co-contraction (Yang & Winter, 1983).

The between-session variability is generally less than the between-subject variability. The SEM/mean ratios range from 0.33 (mean normalised fine-wire EMG of the medial gastrocnemius) and 0.85 (non-normalised fine-wire EMG of the tibialis posterior). The SD/mean ratios range between 0.38 (mean normalised surface EMG of the tibialis anterior and 0.99 (non-normalised fine-wire EMG of the tibialis posterior). The between-session variability may only be slightly much greater than between-subject variability but it suggests that a considerable part of the between-subject variability might be attributable to variability between sessions.

Mean normalisation should thus be considered as the most effective form of normalisation to detect normal from abnormal traces as it can reduce the most between-subject and the between-session variability. Peak normalisation is equally good as there is only small difference between them. Therefore to form a normative EMG template for clinical gait analysis, the between-subject variability should be minimised by peak or mean normalisations. Since persons with neurological problems or certain orthopaedic conditions frequently show abnormal period of activations (De Luca, Kuznetsov, Gilmore, & Roy, 2012; Knutsson & Richards, 1979; Woltering, Guth, & Abbink, 1979), the ideal normative data should facilitate the identification of these abnormalities/differences. Moreover, the peak normalisation would result in the same scaling for reported EMG profiles allowing direct comparisons and clinical interpretations as a percentage of the maximum activity required for the specific task –walking. Therefore, in this PhD work, the peak normalised EMG profiles for these muscles will be used for further analysis of the performance of different sensor types in the next chapter and to form normative dataset.

SD and SEM are the only parameters of variability which have original units and allow direct interpretation. The mean of the SD/mean and mean of the SEM/mean across the gait cycle (Figure 5.5,5.9) clearly show (22-62% between-subject ,0-38 % changes of between-sessions) the effective reduction of variability by mean and peak normalisations and that MVIC is consistently the least effective in reducing variability between healthy participants and often increases variability in comparison to non-normalised data, particularly when using surface sensors. CV (23-85% between-subjects, 5-24% changes of between-session) and VR (11-35% between-subject, 3-13% changes of between-session) seem to be sensitive to normalisation as well. On the other hand, CMC tends to hide the differences (6-22% inter-subject, 2-7% changes of inter-session) (Figure 5.6-8 and 5.10-12). Moreover, the differences between the surface and fine-wire sensors seem less clear with the relative measures of variability.

5.4.4 Limitations

Regarding the EMG measurement of MVIC, it was difficult to ensure maximum participant effort. In this study, verbal encouragement was given during isometric contraction and a visual feedback about the generated force was displayed on the screen next to the participant. Also, the position of the participants was maintained between sessions. Regardless of these efforts, there was still evidence suggesting that MVIC may be another source of errors in grand ensemble average EMG profiles between subjects and between sessions. However, because of the theoretical benefit of this technique, be further investigation to improve its repeatability might prove worthwhile.

5.5 Conclusion and clinical recommendations

There is considerable between-subject variability in non-normalised EMG (up to 100% of the mean signal in the fine-wire EMG of the tibialis posterior and the medial gastrocnemius).

This will make detecting abnormality in clinical practice extremely difficult. Given that being able to distinguish between normal and abnormal is one of the key requirements of any biomechanical measure (Baker, 2006), the potential for normalisation schemes to reduce variability is probably more important than other aspects such as making the measurements more clinically meaningful.

In summary, this study aimed to investigate the effects of different normalisation procedures on the fine-wire and surface EMG of tibialis posterior, tibialis anterior and medial gastrocnemius in healthy adults. Mean normalisation appears to be the best method for reducing variability and this is true across muscles, sensors and different measures of variability. Variability in SD can be reduced by 18% -62% of the mean signal and SEM can be reduced by up to 42% of the mean signal. It has to be appreciated that, through this process, most of the information about the amplitude of the signal is lost and the primary clinical use of the data will thus be to compare the pattern (rather than magnitude of the signal). Mean and peak normalisation (equal performance) should thus be considered as appropriate methods of normalisation to detect normal from abnormal traces.

MVIC normalisation on a dynamometer is generally ineffective in reducing variability and, in many cases, it increases between-subject variability and increased between-session variability of the surface EMG in non-normalised EMG. Also, MVIC may not be possible for persons with neuromuscular disorders (Yang & Winter, 1984) or persons with difficulty communicating.

SEM is the measure that provides the greatest insight into variability and is most sensitive to differences in variability between different normalisation methods. CMC among the selected parameters can be seen to be particularly insensitive to differences in normalisation.

Chapter 6 Application of fine-wire sensors for EMG measurement of tibialis posterior in clinical gait analysis

This chapter aims to address some of the challenges regarding the application of fine-wire sensors in the measurement of the EMG of the tibialis posterior in clinical gait analysis. The tibialis posterior is potentially an important muscle, contributing to gait deviation in participants with neuromuscular deficit, yet little is known about the activation of this muscle during gait. The EMG of the tibialis posterior can only be measured using a fine-wire sensor due to the deep location of the muscle. Furthermore, the use of fine-wire sensors is less common than the use of a surface sensor, as the latter is noninvasive and the process of applying surface electrodes is simpler. In most clinical gait analysis contexts, the fine-wire EMG of the tibialis posterior is collected at the same time as the surface EMG of other superficial muscles but the fine-wire EMG is processed differently (Murley, Buldt, et al., 2009; Murley et al., 2014). It would thus be useful to know whether signals from the two types of muscles (deep and superficial) using different sensors (fine-wire and surface sensors respectively) can be considered as equivalent or whether they actually require different signal processing protocols i.e. the number of gait cycles for the ensemble average and signal filters. Furthermore, because the fine-wire sensor has a small detecting volume, the detected signal may not represent the activity of the entire muscle, so it would be useful to know if the electrode position over/within the muscle has any effect on the signal obtained.

The experiment was designed primarily to compare the signals from fine-wire EMG with those from surface EMG which is more commonly used clinically to detect the activity of muscles. Secondly, the experiment compared the EMG from different sites of fine-wire sensors on the same muscles to test whether the signal can be assumed to be representative of the entire muscle, despite the small detecting volume. The experiment focused on two

muscles whose EMG profiles are particularly well understood, as identified by a systematic review: the tibialis anterior and medial gastrocnemius. Signals from both sensor types were recorded to compare the effect on the patterns and variability of the EMG profiles between subjects and between sessions.

6.1 Background

Sensors play an important role in EMG recording in terms of optimising the signal to noise ratio (Bogey et al., 2000). The sensors should measure only a targeted muscle without signals from the surrounding muscles (*specificity*). The detected EMG should be representative of the entire muscle (*sensitivity*). There are two main types of sensor: surface or skin sensors and inserted (wire and needles) sensors (Basmajian & De Luca, 1985). The surface sensor is the most commonly used, as it is noninvasive, although it is limited to superficial muscles and subjected to crosstalk. The activity of the deep muscles can be recorded by inserted sensors only. The process requires injection by trained and experienced personnel.

Sensors

a) Surface sensors

Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM) defines a *sensor* as ‘ the ensemble of electrodes, electrode construction, and (if applicable) the integrated pre-amplifier’ (Hermens & Merletti, 1999). Surface sensors are widely used because they are non-invasive, cause minimal discomfort and are reasonably reproducible (Jacobson, Gabel, & Brand, 1995) (Figure 6.1). Surface sensors record EMG signals generated from a number of individual motor units within a large volume under the detection surface and are therefore prone to artefacts caused by the movement of the muscle’s innervation zone towards the detection volume during a dynamic contraction (R. A. Bogey,

Perry, Bontrager, & Gronley, 2000; Rainoldi et al., 2000). The signal may also be contaminated by crosstalk from the adjacent muscles (R. A. Bogey, et al., 2000). Reliable results can only be obtained from superficial muscles.

Figure 6.1 a dual-surface electrode

b) Inserted/indwelling sensors

Needle sensors

Needle sensors of various designs incorporate unipolar or bipolar sensors into the hypodermic needle which remains in place for the measurements. Modern manufacturing allows for very small inter-sensor distances and hence highly specific signal detection. They are painful to walk with and highly sensitive to movement, so they are not recommended for use in gait analysis.

Fine-wire sensors

This type of sensor consists of small insulated wires with bared tips (Figure 6.2). They are available as a single wire or as two wires. These wires are threaded through a hypodermic needle for insertion with the tips bent back to form a barb to retain the sensor in the muscle when the needle is withdrawn. The other end of the wires connects to the amplifier. The distance between the bared tips determines the detecting volume. Paired hook wires are inserted together and thus only require a single needle insertion. The tips are only a few

millimetres apart and they thus have a very small measurement volume, so single wire electrodes have to be inserted in pairs requiring two needle insertion. The distance between the tips can be varying depending on the distance between the needle placements but will always be greater than the separation between the tips of the paired wires. They will thus have a significantly larger volume. While the paired hook wires are inserted together, they require only one insertion. Where fine wire sensors are used in gait analysis, the advantage of only requiring one needle insertion in terms of convenience for the assessor and comfort for the patient generally leads to the use of paired wires.

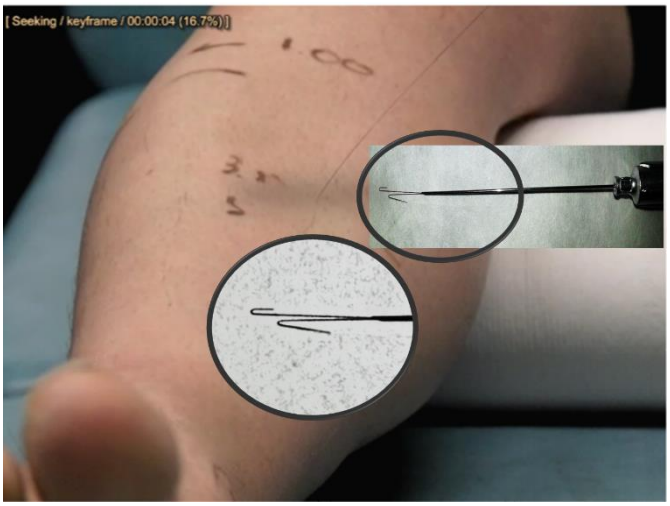


Figure 6.2 Paired hook wires electrode

Although the two types of sensors (surface and fine-wire) have been compared previously, the findings of the studies have been somewhat contradictory. The results of the systematic review in Chapter 3 showed different profiles of rectus femoris. On the other hand, in the study carried out on the vastus medialis and bicep femoris (Jacobson et al., 1995b) a high correlation was found between surface and fine-wire EMG on the vastus medialis and bicep femoris. It appears from the literature that any differences between the surface and fine-wire EMG are muscle-specific.

The comparability between fine-wire and surface EMG is not well understood. In order to determine how best to use fine-wire to record from the tibialis posterior, which is a deep

muscle, the differences between fine-wire and surface EMG should be examined in muscles which can accommodate both and whose activation profiles are well understood i.e. the tibialis anterior and medial gastrocnemius.

As can be seen in Table 6.1, the temporal patterns of EMG profiles in the tibialis anterior, gastrocnemius and soleus between two sensors were similar but the repeatability of fine-wire EMG was less than that for surface EMG. The number of gait cycles included varied between studies and sometimes were not mentioned (Table 6.1). This raises the question of whether the repeatability of fine-wire EMG could be improved using the standard acquisition protocols which address the technical issues related to fine-wire sensors. For example, if the re-test period was also found to be too short [less than two weeks (Paakkari & Mumenthaler, 1974)], this might affect the repeatability between sessions. The measures used to determine the repeatability varied: coefficient of variance (CV), variance ratio (VR) and Pearson correlation coefficient (Bogey et al., 2000; Chimera, Benoit, & Manal, 2009; Kadaba et al., 1985). Since these measures are indicators that do not contain information about the actual signal, the standard deviation (SD) and the standard error of measurements (SEM) are used instead. Our study aims to establish the protocol for fine-wire EMG measurement, so it is important to measure the tibialis anterior and medial gastrocnemius as a reference for validating our protocol against the results from previous studies and determine if different processing is required based on the pattern and variability of the EMG profiles.

Fine-wire sensors are implanted in the muscle of interest and only record signal from a small volume of the muscle so they suffer less from crosstalk. This technique also enables the investigation of deep tissue (Bogey, et al., 2000) and may be minimally affected by the movement of the muscle beneath the skin during a dynamic activity such as gait. However, with a small detecting volume, fine-wire sensors cover fewer motor units, so the detected

fine-wire signal may be more variable than a surface electrode with a larger detecting volume, covering more motor units. The higher variability of fine-wire EMG data may result in a greater number of gait cycles being included to form a representative ensemble average compared to the surface one. This issue will be explored in this chapter where the number of required gait cycles is calculated based on the variability of the EMG profiles between two sensors.

It is known that walking speed may change muscle recruitment in an individual (Den Otter et al., 2004; Hof et al., 2002). Furthermore, walking at different speeds may further affect the muscles, and hence the number of motor units within detection ranges of fine-wire and surface sensors may change, since a surface sensor has a relatively large detecting volume and may detect signals from the adjacent muscles, while a fine-wire sensor has a relatively small detecting volume, and so may be affected differently by various speeds. It is necessary to distinguish the effects on the sensors (system) and the effects on muscle recruitment (actual neuromuscular change). Regarding the fine-wire sensor, there may be a dislocation of the electrode during dynamic movement, so it is necessary to understand the practicality/feasibility of different types of sensor as well as the sensor's performance across different speeds for appropriate selection. Therefore the effect of different speeds on EMG using different types should be explored.

The other issue is that the signal from a small volume may not represent the activity of the entire muscle-sensitivity (Bogey, et al., 2000). The sensor may be able to record from the nearest motor units only. There have, however, been no previous studies to determine the sensitivity of fine-wire signals on the tibialis posterior, tibialis anterior and medial gastrocnemius. A similar study was carried out on the soleus muscle (Bogey et al., 2000). Similar normalised peak and EMG patterns were found from two fine-wire sensors, however, we do not know whether the result would be the same for other muscles as the

motor units may be distributed differently. The authors showed that there was a longer active period of soleus using a surface sensor and concluded that the fine-wire EMG gave more precise signals.

Table 6.1 Performance of surface and fine-wire sensors on muscles acting on the ankle during gait

	Number of participants	Number of included gait cycles	Muscles	Parameters	Finding
Kadaba et al. (1985)	10	3-4	Tibialis anterior and gastrocnemius (rectus femoris, vastus lateralis and semitendinosus were tested along with the ankle muscles)	Variance ratio (VR) (Paired t test), coefficient of variance for median frequency (CV)	Repeatability-The EMG detected by surface sensors are more repeatable between cycles, sessions (on the same day) and days. The median frequency of fine- wire EMG is significantly higher than that of surface EMG.
Bogey et al. (2003)	18	Not clearly stated.	Soleus	Variance ratio (VR)	Mean VR reported by the fine-wire sensor is slightly lower than the surface sensor but this is not significantly different.
Chimera et al. (2009)	11	Not clearly stated.	Lateral gastrocnemius, medial gastrocnemius, soleus and tibialis anterior	Muscle activation onset, time to peak muscular activation and peak amplitude (paired t test) and Pearson correlation coefficient	The EMG profiles from both sensors are similar: correlation coefficient, muscle activation onset and timing of peak. However, peak amplitudes, normalised to maximal voluntary contraction value, detected by surface sensors are higher than fine wire sensors.

In order to determine if fine-wire-EMG data in the tibialis posterior during CGA can be directly compared with surface EMG collected from superficial muscles at the ankle (tibialis anterior and medial gastrocnemius), this study will determine:

1) The effect of sensor types (fine-wire and surface sensors) on variability (between-session and between-session) of EMG profiles from the tibialis anterior and medial gastrocnemius during normal gait;

2) The sensitivity of fine-wire sensors i.e. whether it can provide a representative signal of the entire muscle. The profiles of fine-wire EMG detected by different placements on the tibialis posterior, tibialis anterior, medial gastrocnemius will be compared;
3) The effects of sensor types (fine-wire and surface sensors) on EMG profiles and the practicality/feasibility of measurement at different walking speeds.

The results of this study will be used to determine the collection and analysis procedures for a clinical database of these muscles in practices.

6.2 Research questions

In young healthy adults:

i) How sensitive are fine-wire EMG signals to where the sensors are placed?
ii) How many more gait cycles of fine-wire EMG are required to give the same confidence that the mean is representative compared with surface EMG?
iii) How do EMG signals from surface and fine-wire sensors compare between sessions (repeatability) and across a range of different walking speeds?

Through all 3 questions, both the shape and the variability of the EMG signals will be considered.

6.3 Method

The general methodology for the experiments conducted for this thesis is described in Chapter 4 and Chapter 5 using peak normalisation. Only aspects of the methodology that are specific to this particular study are reported here.

6.3.1 Participants

This study recruited 11 healthy volunteers aged between 18-60 years from among staff and students at the University of Salford, UK.

6.3.2 Data analysis

The mean and standard deviation across a gait cycle are calculated for all EMG profiles. After processing the EMG signals (Chapter 4 and 5), they were presented as a mean or an ensemble average and standard deviation (SD) band to illustrate the dispersion or spread from the average. A low SD indicated that the data points are likely to be very close to the mean. For normally distributed data, one SD on either side of the mean accounts for 68% of the data set. The published gait normative database, which included kinematics and kinetics, was presented in this way as the changes of means and SD over a gait cycle are of interest (Kadaba et al., 1989; Perry, 1992; Schwartz, Trost, & Wervey, 2004; D. A. Winter & Yack, 1987; D.A. Winter, 2009).

The use of linear envelope detection was recommended to represent EMG patterns for dynamic EMG (SENIAM). Ensemble averaging is an important technique to reducing the variability or increasing the repeatability of the EMG profiles during a cyclic movement such as gait (Hermens & Merletti, 1999). SENIAM recommends the user to report the ensemble average and the associated standard error of mean. The standard error of the mean depends on the number of gait cycles being averaged. Therefore, to form the representative patterns, the number of gait cycles included in an ensemble average should be decided. The generally accepted numbers of gait cycles were between 4 and 25 (Kadaba, Wootten, Gainey, & Cochran, 1985; George S. Murley, Buldt, Trump, & Wickham, 2009; George S. Murley, Menz, & Landorf, 2009; Semciw, Pizzari, Murley, & Green, 2013). Therefore, this chapter attempted to calculate the adequate number of gait cycles for processing fine-wire EMG for CGA based on the variability measured from surface EMG, as it was more commonly used in CGA. The peak normalisation was used as it was effective in reducing between-subject variability and between-session variability as described in chapter 5.

a) Comparison of fine-wire EMG detected by proximal and distal sensors

To compare the signals between proximal fine-wire and surface sensors, the Pearson's product correlation coefficient (r) was selected to assess the similarity between the ensemble averaged profiles recorded from both sensors for every 1% interval of gait cycle. The within-subject SD (in session 1) of EMG measured by these sensors for the tibialis anterior and medial gastrocnemius were compared to investigate the variability within subject.

b) Method related to the number of gait cycles

The average within-subject SD, between gait cycles across the gait cycle, for an individual ensemble average, normalised by their peak values, was calculated for the tibialis anterior and medial gastrocnemius for both sensors

In this study the standard error of the mean (SEM_m) is chosen to illustrate the distribution–standard deviation of the sampling distribution of the mean.

$$SEM_m = \frac{SD}{\sqrt{N}}$$

Where SD is the mean standard deviation calculated from 0-100% of the gait cycle and N is the number of included trials

Assumption: as the number of gait cycles used for the surface EMG was studied and recommended to be between 6-8 gait cycles (Shiavi et al., 1998). Therefore the number of gait cycles recorded from fine-wire sensor required to achieve the same level of variability is calculated from the mean SD from both sensors on the same muscle.

$$SEM_F = SEM_S$$

Where: SEM_S = average of the standard error of the mean calculated from the EMG profile using the surface EMG

SEM_F = average of the standard error of the mean calculated from the EMG profile using the fine-wire EMG

$$SEM_F{}^2 = SEM_S{}^2$$

$$\frac{SD_F{}^2}{N_F} = \frac{SD_S{}^2}{N_S}$$

Where: $SD_S{}^2$ is the square of the SD calculated from the EMG profile using the surface EMG

$SD_F{}^2$ is the square of the SD calculated from the EMG profile using the fine-wire EMG

N_S is the number of included trials recorded by surface sensors (6 cycles)

N_F is the number of included trials recorded by fine-wire sensors

Therefore

$$N_F = \frac{SD_F{}^2}{SD_S{}^2} \; x \; N_S$$

c) Comparison between surface and fine-wire EMG

Method related to similarity between surface and fine-wire EMG profiles: The methods were similar to those related to similarity between the surface and fine-wire EMG profiles for the tibialis anterior, medial gastrocnemius and tibialis posterior.

Method related to the repeatability of surface and fine-wire EMG: The standard error of measurement (SEM) of the surface and fine-wire EMG were estimated from the SD and ICC (2,1) which was used as the repeatability of measurement from each individual ensemble average from two sessions. In this model, the participants and measuring sessions were considered to be randomly chosen. The analysis of ICC for this paper was generated using Excel (Microsoft Office 2013, Redmond, WA, USA), and the Real Statistics Resource Pack software (Release 3.5, Copyright 2013 – 2015, Charles Zaiontz) for every 1% interval of gait cycle (100 points). Only the average SEM across the gait cycle was reported for each sensor for comparison.

Method related to the effect of speeds on surface and fine-wire EMG Grand ensemble averages from the participants who completed five different speeds with between-subject SD were presented to demonstrate the normative data from proximal fine-wire and surface sensors, with variability expected between subjects. Then, the similarity of the ensemble averages between changes in the amplitude of the normalised EMG signals and temporal patterns of the profiles caused by different speeds were also investigated using a Pearson correlation coefficient (r). The coefficients were calculated for both sensors.

6.4 Results

In this study, 11 participants: age 33 ±4 years old, 4 females and 7 males, height 1.7 ± 0.07m, and mass 71 ± 12 kg, completed a session walking at a self-selected speed. 10 participants completed two sessions. The average speed for the 11 participants at a self-selected speed was 1.24 ± 0.19 m/s. One participant showed EMG profiles at one session which were so low that there had clearly been a fault with the way in which the measurements were taken or recorded and was therefore excluded. Therefore, the result of the comparison of between-session repeatability in this study was derived from 9 participants: age 33 ±5 years old, 3 females and 6 males, height 1.70 ± 0.07m, and mass 71 ± 13 kg, completed two testing sessions. The average speeds for session one and session two (at least two weeks after the first session) were similar: 1.20 ±0.15 m/s and 1.14±0.17 m/s respectively. The changes in hip, knee and ankle angles during walking were within the normal range implying that the walking patterns for all included gait cycles were similar. At the two fast speeds, the fine-wire sensors were lost in one participant and the motion artefacts were too large to be removed in two participants. Therefore, eight participants were included in EMG profiles across five speeds: five males, three females, averaged age 34±5 years with averaged mass 71±14 kg and averaged height 170± 8 cm.

6.4.1 Comparison of EMG between proximal and distal fine-wire sensors

The correlation coefficient (r) between the proximal and distal fine-wire sensors was high (Table 6.2 and Figure 6.3). The ‘r’ values from all muscles between the proximal and distal sensors were greater than 0.9 and the SD of the proximal sensors from all muscles were slightly less than that of the distal sensors (though this was slight compared with the variability between participants [Table 6.3]). From visual inspection, there was no difference of the onset of muscular activation, the timing of peak activation or normalised

peak amplitude between two placements of sensors in any of the three muscles. In summary, the placement of the fine-wire electrodes only has a small effect on the signal record. In the presentation of further results and the discussion below, only the proximal fine-wire sensor will be the focus.

Table 6.2 Comparison of EMG profiles from proximal and distal fine-wire sensors for the *tibialis posterior*, *tibialis anterior and medial gastrocnemius. The average SD* is the within session SD calculated at 1% intervals of the gait cycle and then averaged across the gait cycle. The Pearson correlation coefficient is that between the within session average EMG signals from both sensor types across the gait cycle.

	Tibialis posterior			*Tibialis anterior*			*Medial gastrocnemius*		
ID	Average SD (%)		Pearson correlation coefficient (r)	Average SD (%)		Pearson correlation coefficient (r)	Average SD (%)		Pearson correlation coefficient (r)
	Proximal	Distal		Proximal	Distal		Proximal	Distal	
1	8	10	0.989	10	8	0.974	7	7	0.994
2	7	10	0.979	8	6	0.986	9	9	0.987
3	10	16	0.953	11	9	0.980	9	9	0.983
4	12	17	0.947	16	12	0.946	8	8	0.996
5	10	12	0.922	10	12	0.961	6	6	0.991
6	14	12	0.940	8	7	0.988	7	6	0.998
7	11	11	0.986	8	11	0.970	5	9	0.955
8	9	10	0.954	7	9	0.994	8	7	0.994
9	15	11	0.871	10	8	0.996	7	7	0.960
10	20	17	0.917	15	14	0.973	4	19	0.967
11	9	8	0.992	6	6	0.964	7	6	0.992
Average	11	12	0.950	10	9	0.975	7	8	0.983

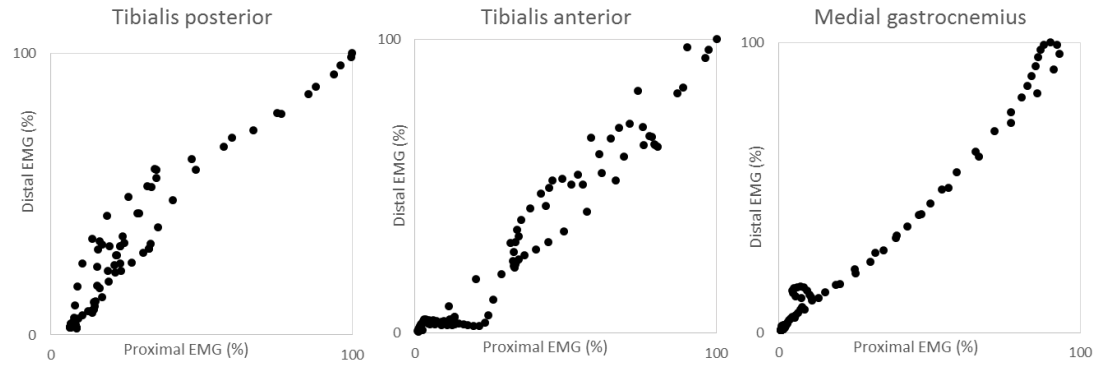


Figure 6.3 Scatter plots of the proximal and distal fine-wire EMG *Tibialis posterior*, *tibialis anterior* and *medial gastrocnemius* scatter plots are based on the ensemble averages of subject with middle-ranking r-value, subject 3, 1 and 5 respectively ($p<0.001$)

Table 6.3 Between-subject standard deviation

	Between-subject standard deviation (SD) (% of peak)		
	Range	Median	Average
Tibialis posterior			
Proximal fine-wire	3-33	10	12
Distal fine-wire	2-25	9	10
Tibialis anterior			
Proximal fine-wire	2-22	14	14
Distal fine-wire	2-23	15	14
Surface	4-27	14	14
Medial gastrocnemius			
Proximal fine-wire	1-28	8	11
Distal fine-wire	4-30	10	13
Surface	1-29	7	10

6.4.2 Number of gait cycles

Within the tibialis anterior and medial gastrocnemius muscles, the difference the grand ensemble averaged SD across the gait cycle was small for the two types of sensor and the average across the gait cycle is thus given in Table 6.4. Although the standard deviations for the gastrocnemius are generally lower than those for the tibialis anterior, normalization to the peak values means that the two are not directly comparable. The differences between the standard deviations from the different sensor types are small compared to those between individuals. The number of gait cycles required to give a SEM_m for fine-wire electrodes equivalent to that obtained from 6 gait cycles of surface EMG for the tibialis anterior ranged between four and eight, with a mean of five gait cycles. For the medial gastrocnemius, the number ranged between two and eight, with a mean of six gait cycles. The overall conclusion is that the within session repeatability for both sensor types is similar and thus that a similar number of gait cycles is required in order to have confidence in the mean signal regardless of sensor type. Therefore, for further analysis, six trials were obtained from every sensor to form an individual ensemble average.

Table 6.4 Comparison of fine-wire and surface sensors for the tibialis anterior and medial gastrocnemius. The s*verage SD* is the within session SD calculated at 2% intevals of the gait cycle and then averaged across the gait cycle. *Number of gait cycles* is the number of gait cycles of fine-wire EMG data that must be combined to give the same standard error of measurement as 6 gait cycles of surface EMG data. The *Pearson correlation coefficient* is that between the within session average EMG signals from both sensor types across the gait cycle.

	Tibialis anterior				*Medial gastrocnemius*			
ID	Average SD (%)		Number of gait cycles	Pearson correlation coefficient (r)	Average SD (%)		Number of gait cycles	Pearson correlation coefficient (r)
	Fine-wire	Surface			Fine-wire	Surface		
1	10	13	4	0.922	7	7	6	0.994
2	8	10	4	0.980	9	11	4	0.993
3	11	13	4	0.973	9	6	6	0.993
4	16	14	8	0.976	9	9	5	0.996
5	10	10	6	0.948	6	5	9	0.993
6	8	8	6	0.986	7	7	6	0.992
7	8	9	5	0.986	5	5	6	0.978
8	7	8	5	0.966	8	6	7	0.988
9	10	9	7	0.991	7	7	8	0.987
10	15	18	4	0.966	7	6	2	0.933
11	6	6	5	0.954	7	7	6	0.988
Average	10	11	5	0.968	7	7	6	0.985

6.4.3 Comparison of the EMG profiles between surface and fine-wire EMG

Similarity between surface and fine-wire EMG profiles

The correlation coefficient (r) between two types of sensor for the tibialis anterior and medial gastrocnemius muscles was higher than 0.9 for each participant (0.922-0.996) (Figure 6.4 and Table 6.4). However, between 30% and 50% of the gait cycle, there was a slight difference in activation pattern detected by surface and fine-wire EMG in the tibialis anterior. The surface sensors detected continuous activity but it was relatively quiet in profiles detected by fine-wire sensors (Figure 6.5).

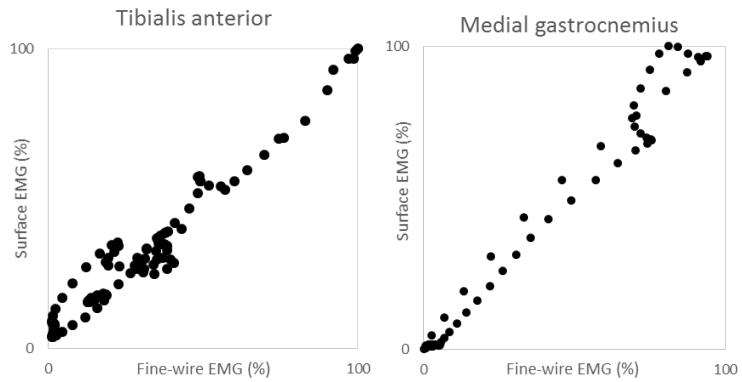


Figure 6.4 Scatter plots of fine-wire and surface EMG. The *tibialis anterior* and *medial gastrocnemius* scatter plots are based on the ensemble averages of subjects with a middle-ranking r-value, subject 3 and subject 6 respectively. ($p<0.001$)

Repeatability of surface and fine-wire EMG

Table 6.5 shows that the SEM of EMG detected by the surface sensors was less than those of the fine-wire sensors. Within the tibialis anterior, the fine-wire sensors showed a higher average SEM than the surface sensors, which also had a narrower range. The difference in the averaged SEM and the ranges between the sensors was smaller for the medial gastrocnemius. Generally, SEM is large when the muscle is active, (particularly at peak) and minimal when the muscle is inactive during the gait cycle (Figure 6.5).

Table 6.5 Standard error of measurement of the EMG profiles across the gait cycle

	Standard error of measurements of EMG between sessions (%)		
	Range	Median	Average
Tibialis posterior (fine-wire)	3-32	14	14
Tibialis anterior (fine-wire)	2-18	11	10
Tibialis anterior (surface)	1-10	5	4
Medial gastrocnemius (fine-wire)	2-19	8	9
Medial gastrocnemius (surface)	1-21	6	7

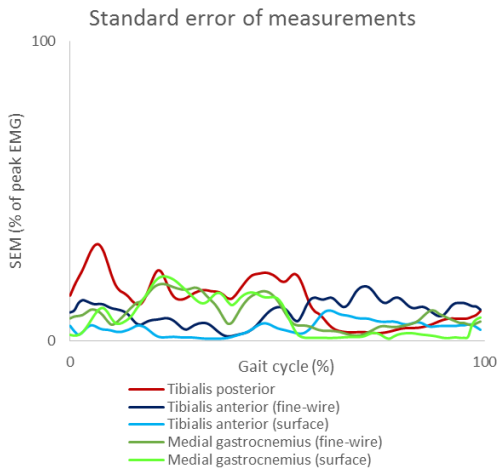


Figure 6.5 Standard error of measurement across the gait cycle

Effect of speeds on surface and fine-wire EMG

The detected averaged speeds differed by 25% of self-selected speeds as instruction (Table 6.6). The temporal patterns of the EMG profiles for the tibialis anterior and medial gastrocnemius across different speeds were remarkably consistent in regard to both the average patterns recorded and the variability around them, suggesting that both sensors are recording the same thing (Figure 6.6-9).

Table 6.6 Speeds

	Slowest	*Slower*	*Self-selected*	*Faster*	*Fastest*
Speed (m/s)(SD)	0.64(0.15)	0.94 (0.14)	1.18 (0.15)	1.46 (0.19)	1.86 (0.24)
Toe-off (%)(SD)	62 (2)	61(2)	60 (1)	61 (3)	58 (3)

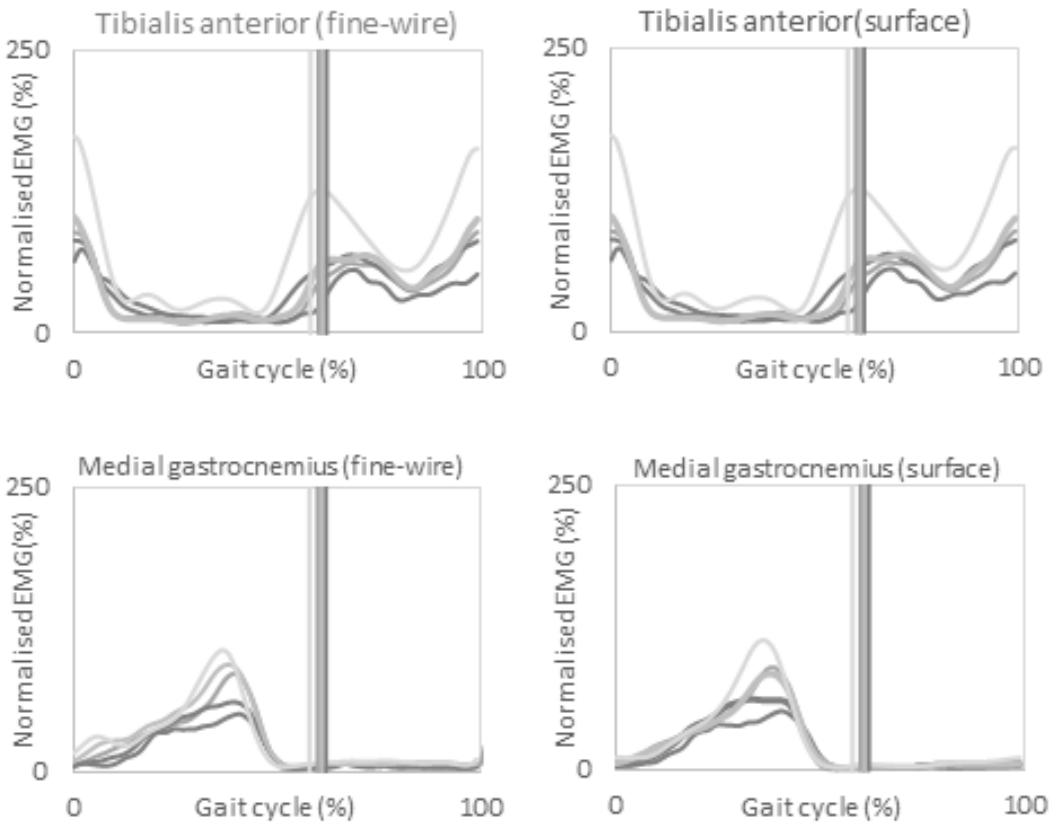


Figure 6.6 Grand average of EMG profiles across 5 speeds from eight subjects. The darkest line is from the slowest speed and the lightest line is from the fastest speed. (The y-axis is the EMG normalised to the peak value at the self-selected speed)

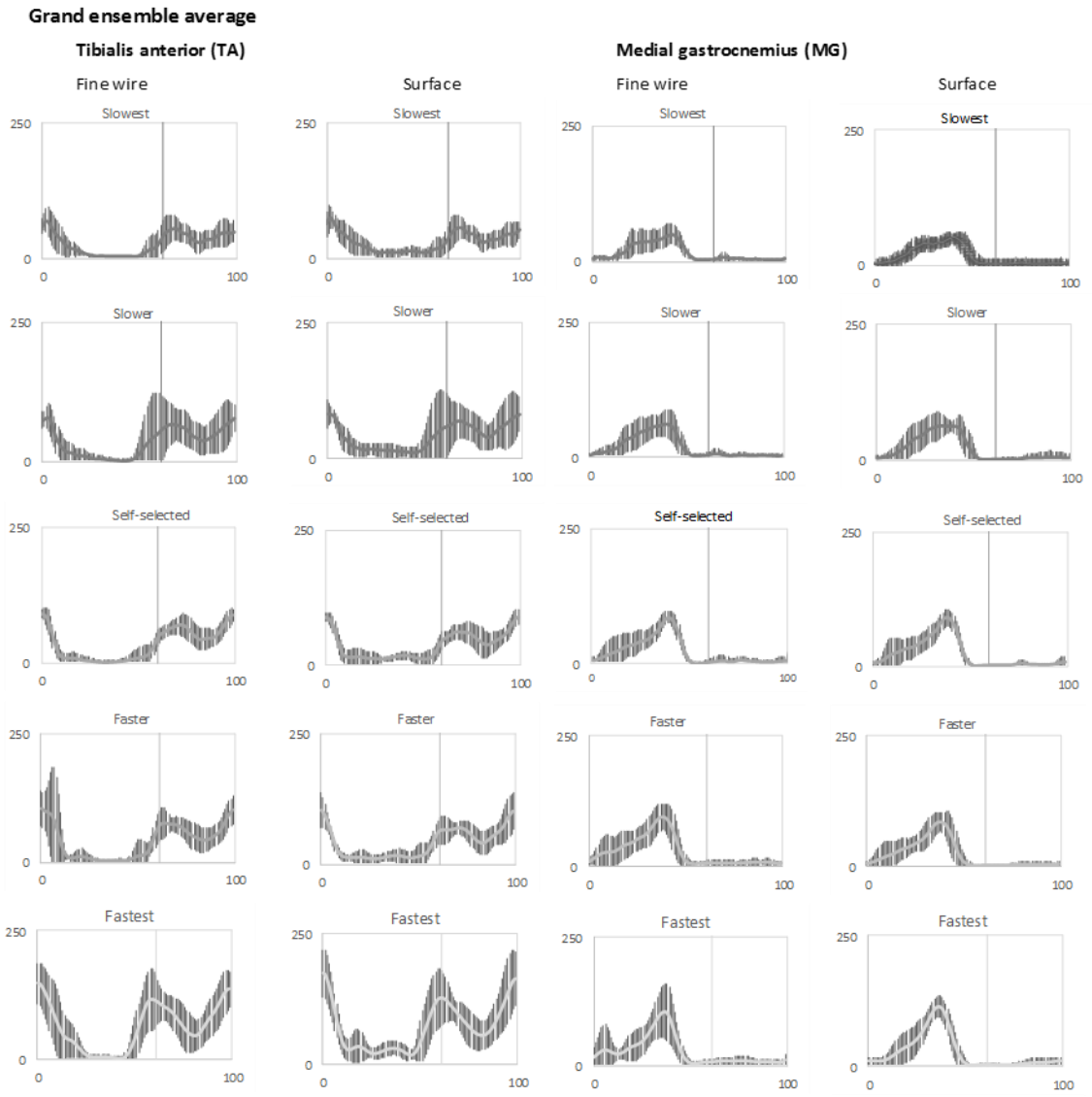


Figure 6.7 Grand ensemble average of EMG profiles with SD across five speeds from eight subjects (The x-axis is the percentage of gait cycle and the y-axis is the EMG normalised to the peak value at a self-selected speed)

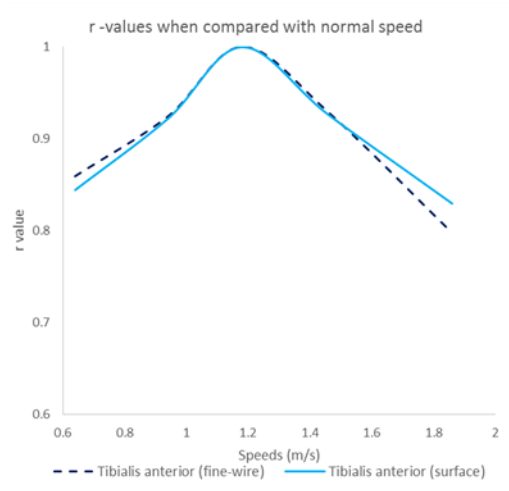


Figure 6.8 Median correlation coefficient (r-value) between fine-wire and surface EMG of the tibialis anterior across different speeds

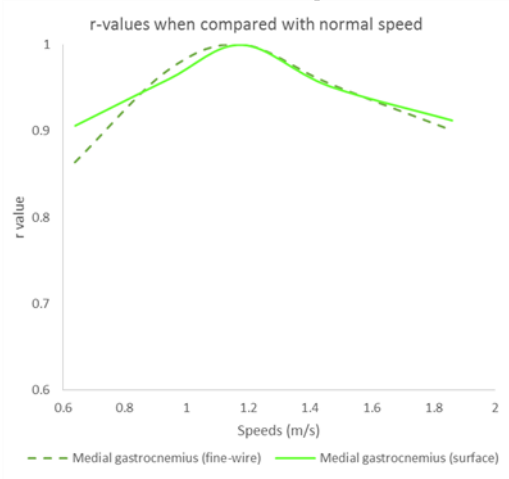


Figure 6.9 Median correlation coefficient (r-value) between fine-wire and surface EMG of the medial gastrocnemius across different speeds

6.5 Discussion

6.5.1 Comparison of EMG between proximal and distal fine-wire sensors

The normalised ensemble averaged EMG profiles detected by the proximal and distal fine-wire sensors are similar in the tibialis anterior, medial gastrocnemius and tibialis posterior. The slight difference in EMG amplitude may be the result of the muscular architecture. It is not homogenous as there is a difference in the density of the collagenous sheaths through the muscles (Perry, Easterday, & Antonelli, 1981). In this study, the difference in amplitude is accommodated to a certain level using the peak normalisation technique. High correlations between normalised EMG profiles of proximally and distally located fine-wire sensors indicate the similarity of the patterns (Table 6.2). For all participants, the timings of peak activation are identical between the two sensor locations during the same session. Similarity in the pattern, variability and timing of the peaks implies that, within the detectable area of the tibialis posterior, medial gastrocnemius and tibialis anterior, fine-wire sensors can detect signals from a similar number of motor units, similar fibre types, muscle fibre architectural characteristics and neuromuscular partitioning, as these anatomical factors affect the EMG signals (Romkes & Brunner, 2007). The reason for this similarity may be because of the similar dispersion of the motor units within these testing volumes. Bodine-Fowler et al. found the statistical similarity of muscle fibre types throughout the muscle volume in cats which could be expected in other mammals such as humans (Bodine-Fowler, Garfinkel, Roy, & Edgerton, 1990). Another study also found a similar arrangement of muscle fibres throughout the vastus lateralis in males (Lexell & Downham, 1991). Therefore, although the fine-wire sensor may only detect action potential from a small volume, the signal detected by the fine-wire sensor can represent the activity of the entire muscle for the tibialis posterior, tibialis anterior and medial gastrocnemius using our protocol. As there was no theoretical justification for the selection of the proximal or distal fine wire sensors, as they

were similar in pattern and variability, only proximal fine-wire EMG was used for the other analyses in our study.

6.5.2 Number of gait cycles

SENIAM recommended that users report the ensemble average signal and the associated standard error of mean to indicate the variability of the reported EMG profiles (Hermens & Merletti, 1999). The standard error of the mean (variability) of an individual ensemble average is influenced by the number of gait cycles being averaged. Therefore, the number of gait cycles should be controlled. For surface EMG, two articles reported the results from the experiment when different numbers of gait cycles were used (Arsenault et al., 1986; Shiavi et al., 1998). Arsenault et al. (1986) concluded that the variability obtained from three gait cycles was similar to that from 12 gait cycles using the analysis of variance, while Shiavi et al. (1998) illustrated that six to eight gait cycles were required to reduce variability based on the normalised mean square error. Combining these studies suggests that a minimum of six gait cycles is required to give confidence in an ensemble averaged surface EMG signal. No such guidelines regarding the number of gait cycles required for fine-wire EMG have been identified. Our experiment showed that to achieve the same level of standard error of mean as the ensemble average of surface EMG (N=6), a similar number of six gait cycles were required from fine-wire EMG for the tibialis anterior and medial gastrocnemius. This is because the standard deviation for the fine-wire signals is similar to that for surface EMG signals. Therefore for the tibialis posterior, where the surface EMG is not applicable, six gait cycles are recommended and will be used for all future collections within this work.

The differences between the average within-subject variability (in the same session) indicated by the SD calculated from the tibialis anterior (10% for fine-wire and 11% for

surface) and medial gastrocnemius (7% for fine-wire and surface) between both types of sensors were small.

The within session, within-subject variability of fine-wire EMG is similar to surface EMG. For the medial gastrocnemius, the similar within-subject variability between surface and fine-wire EMG contradicted the result of Kadaba et al.(1985) as they concluded that the variation of fine-wire EMG was higher than the surface as fine-wire may cause intramuscular bleeding, possible displacement, deformation and fracture of the electrode (Kadaba et al., 1985). However, later study showed that the insertion of fine-wire within the muscles did not alter the activation patterns of the muscles (Jacobson et al., 1995a). Also, the chance of fine-wire migration during testing can be reduced by allowing a minimum of six strong contractions of the muscles prior to the data collection to ensure the secured positions of the wire within the muscle (Perry & Bekey, 1981). In approximately 50% of participants in this study, the within session, within-subject variability of the fine-wire EMG was slightly less than the surface EMG in the tibialis anterior, probably due to the flexibility of the fine-wire and its position within the muscles, so when there were geometric changes during muscle contraction, the signal recorded was less affected by this than the surface EMG (Chapman et al., 2010; Chimera et al., 2009). Kadaba's produced a similar finding for the tibialis anterior (the variance ratio of fine-wire was 0.17 and that of surface was 0.23).

6.5.3 Comparison of the EMG profiles between surface and fine-wire EMG

a) Similarity between surface and fine-wire EMG profiles

The temporal patterns of the EMG profiles from the tibialis anterior and medial gastrocnemius detected by both sensors are similar, as indicated by the high correlation coefficient. The patterns of the tibialis anterior in medial gastrocnemius profiles measured in our experiment are also similar to the profiles shown in the systematic review (Chapter

3). In a similar study, Chimera et al. (2009) also showed a high 'r': 0.94 for medial gastrocnemius and 0.85 for tibialis anterior. Therefore EMG signals from different sensors are comparable using our protocols. There is no particular reason to expect the tibialis posterior to act differently and it seems reasonable to conclude that fine-wire EMG from the tibialis posterior can be considered as equivalent to surface EMG from other muscles in clinical gait analysis (provided that the protocols used for collecting and processing both are as described in the methodology section of this chapter).

Fine-wire and surface sensors have between-subject variability (between-subject SD) of below 20% of the peak amplitude (tibialis anterior 13%, 14.5%; medial gastrocnemius, 11.5% and 10.9%, respectively). This level of between-subject variability is well below the SD pooled from different studies, as shown in Chapter 3 Systematic review (tibialis anterior 24% and medial gastrocnemius 25%). This level of variability should allow the detection of abnormality. The smaller SD calculated in our data may be because of the homogenous data acquisition and processing. The differences in between-subject variability between two sensor types are small and considered to be not clinically significant using our protocol.

Despite the high correlation coefficient and similar within session, within-subject and between-subjects SD between the two sensor types, a fine-wire sensor tends to be slightly more specific than a surface sensor. Regarding the tibialis anterior, there is a consistent difference between two detected EMG profiles during a period between 30-50% of gait cycle, when the fine-wire signal shows a quiet period, while the surface signal shows a low degree of activity among the muscles. The appearance of continuous activity detected by the surface sensor is found in Chimera's study (2009) during gait. This activity occurs at approximately the same period as peaks in the plantarflexors: medial gastrocnemius and tibialis posterior in our study, so this difference may be crosstalk from the plantarflexors including the soleus which is close to the tibialis anterior but in the posterior compartment.

As the activity of peroneus longus was not collected, it is difficult to determine whether this influences this additional signal. Solomonow et al. demonstrated that crosstalk may originate from plantarflexors as they found a linear relationship between crosstalk in the tibialis anterior and lateral gastrocnemius, and the EMG of medial gastrocnemius at different level of motor unit recruitment (Solomonow et al., 1994). In addition to the tibialis anterior, the fine-wire EMG profiles of rectus femoris from Barr et al (2010) and Nene et al. (2004) showed rectus femoris activity during the transition from stance to swing only and illustrated that the other burst of activity is crosstalk from vastus lateralis. The different patterns detected by the two sensor types may be caused by the different detecting volume. Although previous authors (Barr et al., 2010; Chimera et al., 2009; Nene et al., 2004; Solomonow et al., 1994) have suggested that different sensor types might have different degree of sensitivity to cross-talk, our data suggest that this is quite a small effect for the muscles studied and unlikely to be clinically significant.

b)Repeatability of surface and fine-wire EMG

The result from this study shows similar SEM from fine-wire sensors compared with surface sensors (2-6% difference) suggesting that the EMG signals obtained from surface sensors is as repeatable as those detected by fine-wire sensor between sessions. The difference between the self-selected speeds between two sessions is small (the difference is 0.05 m/s, which is less than the SD), implying that the change in EMG signal is not caused by speed difference. The variability between the profiles of fine-wire and surface EMG are assumed to be from the re-application of sensors and changes in individual conditions. Similarly, Kadaba et al. (1985) also found that the averaged variance ratio (VR) calculated from fine-wire signals was similar to that from the surface for both muscles (tibialis anterior: 0.51 (fine-wire), 0.48 (surface); medial gastrocnemius: 0.65(fine-wire), 0.58 (surface)), though their VR was higher than our result (Chapter 5). When there is a re-application of the sensor in different

sessions, the slight displacement influences the signals detected from the local motor units. Our between sessions, within-subject variance ratio reported in the normalisation chapter are lower than those reported by Kadaba et al (1985). Bogey et al. (2003) reported the repeatability of the soleus with the averaged VR 0.19 for fine-wire sensors which was similar to that for the surface sensor (0.20). In our study, according to the tibialis anterior, the between sessions, intra-subject VR from fine-wire is slightly less than the VR from the surface sensors but the VR from the fine-wire sensor of the medial gastrocnemius is still slightly higher than the surface sensors. However, the SEM of fine-wire is slightly higher than that of the surface in the tibialis anterior (6%) and medial gastrocnemius (2%). The differences are small and whether one is higher or lower than the other is largely a matter of chance. Although it has been suggested that fine-wire sensors may be more sensitive to position, leading to greater between-session variability, our data confirm Kadaba's suggestion that this effect is small, probably too small to be clinically significant.

The difference of the between session, within-subject variability between Kadaba's study and the others (Bogey et al. (2003) and our study) may be because of the different intervals between test days (7 days in Kadaba et al.) and they calculated VR from 4 days, while Bogey et al. set the second session after approximately 17 days and our study allowed a minimum of 14 days to alleviate intramuscular hematoma and disruption of the muscle tissue (Paakkari & Mumenthaler, 1974). Needle myopathy may disrupt the conductivity of the electrical signals and, therefore, influence the repeatability of fine-wire EMG (Chapman et al., 2010). As shown, the between session, within- subject variability of surface and fine-wire sensors are slightly different and maybe negligible for the tibialis anterior and medial gastrocnemius, provided that appropriate data acquisition is applied. Therefore, using standard protocols, both sensors are sufficiently reliable to be used for repeated measures of the tibialis anterior and medial gastrocnemius during gait at self-selected speeds.

c) Effect of speed on surface and fine-wire EMG

The similarity between fine-wire and surface EMG is high when walking speeds are within 50% faster or slower than self-selected speeds. The faster speeds tend to increase the EMG amplitude and early timing of the peak in the tibialis anterior and medial gastrocnemius, regardless of sensor type. At an individual level, when the walking speed increases, the similarity of the EMG between two sensor types decreases but only slightly. When the walking speed is 50% faster than normal, there was a report of fine-wire dislocation and excessive noise in fine-wire EMG. This resulted in data missing for 2 participants. Therefore, in fast dynamic tasks, fine-wire sensors may not be an appropriate choice.

When the speed increases, EMG profiles from both sensors a show higher amplitude and early timing of the peak. The changes in the average profile detected by fine-wire sensors were more systematic than surface EMG across different speeds, but the average surface EMG profiles detected sometimes overlapped. The less distinguishable changes in the surface EMG may be because the surface sensor has a larger detecting volume than the fine-wire sensor, thereby making the surface EMG less sensitive to small changes. The surface EMG of the medial gastrocnemius in a previous study on children showed a systematic change in the amplitude and timing of peaks using a more sophisticated speed categorisation (Schwartz et al., 2008). Our data suggest that fine-wire EMG has a tendency to be more sensitive to speed changes (± 25% and ±50% of self-selected speed) than surface EMG. However, these differences are small and may not be clinically important.

The within session, between-subject variability (average SD) of fine-wire and surface sensors altered in a similar manner across the speeds but the differences are small and there is no clear consistent pattern. The reason may be the difference in motor control or effort required by different individuals to achieve the target speeds, so the application of different

sensors is unlikely to affect the variability of the grand EMG profiles of the tibialis anterior and medial gastrocnemius.

6.5.4 Limitations

This chapter set out to compare the different sensor types on the EMG measurement of the tibialis anterior and medial gastrocnemius. It would have been ideal to measure them on the same site of the muscle, but this was impractical in our experimental setting as the use of fine-wire may cause heamatoma and may alter the signal detected by the surface sensor. Therefore, the guideline for the surface sensor was strictly followed and the fine-wires were inserted as close as possible without being affected by the bandage wrapping around the surface electrode. The fine-wire electrodes were applied to the detectable area, far from the tendon. This was confirmed by the ultrasound images (Chapter 4).

The use of the 50 Hz high pass filter to process surface EMG is higher than the recommended 20Hz filter (De Luca, Gilmore, Kuznetsov, & Roy, 2010). However, the difference required to alter the pattern and timing of the peak of the detected linear envelope is small, so it is important to keep the processing between the surface and fine-wire EMG identical to allow direct comparison between these signals.

6.6 Conclusion

Regarding the research questions:

i) How sensitive are fine-wire EMG signals to where the sensors are placed?

Our data showed that fine-wire EMG signals are not sensitive to where the sensors is placed in muscle belly based on the similarity of waveform and variability of EMG profiles.

ii) How many more gait cycles of fine-wire EMG are required to give the same confidences that the mean is representative compared with surface EMG?

Minimum of six gait cycles are required.

iii) How do EMG signals from surface and fine-wire EMG sensors compare between sessions (repeatability) and across a range of different walking speeds?

Surface and fine-wire EMG have similar level of repeatability and waveforms between sessions and across a range of walking speeds.

Chapter 7 Incorporating EMG data with kinematic and kinetic data to understand walking in healthy adults

The previous chapters have described and tested protocols for capturing and processing both surface and fine-wire EMG in a manner that is comparable with kinematic and kinetic data. This will be illustrated by presenting data from the medial gastrocnemius, tibialis anterior and tibialis posterior in association with kinetic and kinematic outputs of the conventional gait model. In this chapter, this will be used to produce a better understanding of healthy walking. In the following chapter, it will be applied to a case series of three stroke participants.

7. 1 Background

In CGA, patients with post-stroke or other neuromuscular disease may walk with gait deviation, identified by kinetics and kinematics, at a wide range of speeds. EMG data allow clinicians to observe muscle dysfunction which may contribute to gait deviation. In order to differentiate the pathological features from those that change due to walking at different speeds, it is necessary to provide simultaneous EMG, kinematics and kinetics.

Two particular issues facing clinical gait analysis are different self-selected walking speeds and ages in patient groups compared to the normative reference. It is thus necessary to understand how the gait pattern changes with different walking speeds. For example, the reported walking speeds of children with cerebral palsy ranges from 0.13-1.45 m/s (Schwartz, Viehweger, Stout, Novacheck, & Gage, 2004). Also, older adults tend to walk more slowly than younger adults. With particular regard to EMG, there is a further question of whether age itself affects the measurements.

7.1.1 Effect of different speeds

Different speeds are known to affect the kinematics of walking (e.g. Kirtley, Whittle, and Jefferson (1985); Schwartz et al. (2008); Shiavi et al. (1987a)) and muscle recruitment controlling forces affect the progression of the body segments (Den Otter et al., 2004; Nymark et al., 2005; Schwartz et al., 2008; Yang & Winter, 1985). Therefore different speeds are likely to affect EMG, as has been observed experimentally in a number of studies (Schwartz et al., 2008; Stoquart, Detrembleur, & Lejeune, 2008; Van Hedel, Tomatis, & Muller, 2006). Generally, the amplitude of lower limb EMG profiles increases as the speed increases due to the higher muscular force demand (Den Otter et al., 2004). The change is muscle specific (Hof et al., 2002; Yang & Winter, 1985).

Tibialis anterior and medial gastrocnemius activation profiles across different speeds have been reported more extensively than for the tibialis posterior (e.g. Clancy et al. (2004); Den Otter et al. (2004); Nymark et al. (2005)). It was found that a slow speed altered the timing, and decreased the magnitude of the EMG signals (speed range 0.2-2.1 m/s) (Clancy et al., 2004; Nymark et al., 2005; Warren et al., 2004). Hof et al. (2002) proposed functional groups/equations to estimate the EMG for each muscle at any speed from a constant and a proportionally increasing factor. The previous dataset lack reports of the activity of the deep muscle tibialis posterior, frontal plane kinematics and kinetics at the ankle joint across gait cycles with a standard deviation which indicates variation between healthy participants (Schwartz et al., 2008; Stoquart et al., 2008; Van Hedel et al., 2006).

Only two articles reported collections of tibialis posterior activation over a range of speeds: Kameyama et al. failed to report the tibialis posterior but raw EMG of the lateral gastrocnemius from a single gait cycle because they had similar patterns at self-selected speeds and Murley et al. reported tibialis posterior profiles for 30 young adults (24 ± 6 years) who were restricted to a neutral foot posture (Figure 7.1); none of these studies reported

simultaneous kinetics and kinematics (Kameyama, Ogawa, Okamoto, & Kumamoto, 1990; Murley et al., 2014). It is necessary to obtain normative EMG data from the tibialis posterior in participants without being constrained to a specific foot posture in pain-free, healthy adults.

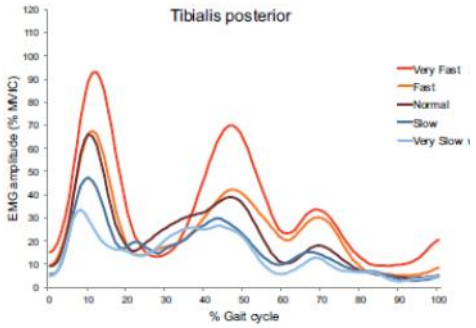


Figure 7.1 Grand EMG ensemble average of the tibialis posterior derived from a single gait cycle for each participant across five speeds (Murley et al., 2014)

7.1.2 Effect of age

Age is also an important factor influencing walking performance. In order to underpin the use of EMG within CGA, a comparison of patient data to age-matched cohorts should be considered as many patients are in the older age group. Benjamin, Qin, and Ralphs (1995) reported that an approximately 49% incidence of stroke occurred in adults aged over 54 years. At the level of muscle tissue, with aging, there is a remodelling of the motor system, including changes in muscle properties and neural pathways (Enoka, 2008). The remodelling of the motor system includes decreases in the number of functional motor units (Campbell, Mccomas, & Petito, 1973), alterations in the motor unit territories, a decrease in the average frequency of the motor unit action potentials (MUAP) and changes in the muscle control mechanisms (Enoka, 2008; Erim, Beg, Burke, & De Luca, 1999). The remodelling of the motor system with age is therefore likely to decrease the EMG signals: amplitude and frequency.

Sarcopenia is a decrease in muscle mass and strength (Evans, 1995). Several studies report decreases in a cross-sectional area of the thigh muscle (Klitgaard et al., 1990; Lexell, Taylor, & Sjöström, 1988) and reductions in leg muscle mass (Janssen, Heymsfield, Wang, & Ross, 2000). These may be due to a loss of muscle fibres deprived of innervation, muscle fibre atrophy and fibre type grouping (Nilwik et al., 2013). The proportion of muscle type I occupied in muscle volume increases as there is atrophy of muscle Type II (Klein, Marsh, Petrella, & Rice, 2003; Nilwik et al., 2013). Therefore, muscles have slower contractile properties (D'antona et al., 2003), possibly leading to stiff joints (Schmitz, Silder, Heiderscheit, Mahoney, & Thelen, 2009). Both of these changes may have consequences for EMG, kinematics and kinetics, measured during CGA.

There are few EMG studies aimed at describing the EMG profiles in older adults (>50 years) (Chung & Giuliani, 1997; Schmitz et al., 2009) compared to those focusing on younger adults (20-40 years). EMG data from healthy, older adults is more frequently collected as a control group for a patient cohort and there are variations in the collection and analysis protocols between studies [e.g. (Al-Zahrani & Bakheit, 2002; Childs, Sparto, Fitzgerald, Bizzini, & Irrgang, 2004; Lynn & Costigan, 2008)]. Studies that provide the EMG of the medial gastrocnemius and tibialis anterior in older adults show an increase in medial gastrocnemius activity during 0-10% of the gait cycle and in tibialis anterior activity during 10-30% of the gait cycle compared to younger adults (Schmitz et al., 2009).

Compared to EMG, there are many studies reporting changes in kinetics, kinematics and spatial temporal parameters due to aging. Older participants are likely to have slower walking speeds with short step lengths, higher variability in step length, and longer support time and step width (Chung & Giuliani, 1997; Oberg, Karsznia, & Oberg, 1993). There are reports of reductions in ankle power (Mcgibbon & Krebs, 2004; Winter, 1991) and the range of plantarflexion (Kaneko, Morimoto, Kimura, Fuchimoto, & Fuchimoto, 1991). Despite the

several reports of changes in gait variables with the aging process, the underlying changes that occur remain unclear (Mcgibbon, 2003; Schmitz et al., 2009). It is likely that the neuromuscular changes in aging partly contribute to the kinematic and kinetic changes and EMG provides a window into these neuromuscular processes. If there are fewer motor units firing at a lower frequency in older adults, then EMG is likely to have lower amplitude and lower power at higher frequencies. This may coincide with a reduced power and range of motion when force production decreases as a result of neuromuscular changes.

Previous studies have compared either kinematics/kinetics or EMG in superficial lower limb muscles without addressing the important deep muscle tibialis posterior (e.g. Chung and Wang (2010); Chung and Giuliani (1997); Kang and Dingwell (2008); Schmitz et al. (2009). Murley et al. did not include data on older adults (Murley, Buldt, et al., 2009; Murley et al., 2014; Murley, Menz, & Landorf, 2009). Barn et al. compared the EMG of the tibialis posterior, joint kinematics and kinetics (only during the stance phase) between patients and aged-match healthy, older adults, but did not compare younger and older adults (Barn et al., 2014). Studies comparing the EMG of the tibialis posterior in healthy younger and older adults have not, to the author's knowledge, been found.

In order to provide a normative fine-wire EMG dataset of the tibialis posterior for CGA use, other muscles around the ankle with surface EMG profiles (tibialis anterior and medial gastrocnemius) will also be collected along with kinetics and kinematics to allow the identification of muscle dysfunctions which are likely to cause pathological gait deviation. Also, the EMG profiles across different speeds are reported to allow the detection of changes due purely to different walking speeds from those due to pathology. Secondly, the EMG profiles of the tibialis posterior in older adults are collected to identify the changes, caused by aging, manifest in the activation profiles of this muscle during gait as a guide to determine

if a normative EMG data for the older age group is necessary to aid the clinical assessment of gait deviation.

7.2 Research questions

The primary research question is how EMG data from both surface and fine-wire can be integrated more effectively within the conventional methods of clinical gait analysis in healthy adults. This will be illustrated by integrating EMG data from the lower leg muscles in healthy adults to answer two subsidiary questions:

i) How does walking speed affect the activity of the muscles of the lower leg and associated kinematics and kinetics?

ii) Is there any evidence that the neuromuscular changes associated with ageing are manifested in the EMG of the lower leg muscles during walking?

7.3 Method

Data were collected and processed using the protocols outlined in Chapter 4 specific to the proximal fine-wire sensor for the tibialis posterior and surface sensors for the tibialis anterior and the medial gastrocnemius. The procedures for peak normalisation are described in Chapter 5. The following sections are specific to this chapter:

7.3.1 Participants

This study recruited eleven healthy volunteers aged between 18-50 years old and four healthy participants who were aged over 50 years old from the University of Salford, Salford, UK.

7.3.2 Protocol

After completion of the marker placement, participants in the younger groups were asked to walk along a 6m walkway in the Gait Laboratory at 5 different speeds: self-selected speed,

a 25% slower speed, a 50% slower speed, a 25% faster speed, and a 50% faster speed. The four participants in the older age group were asked to walk at a self-selected speed only.

7.3.3 Data analysis

The mean and standard deviation across a gait cycle was calculated for all EMG profiles. The processed EMG signals were presented as an ensemble average with a standard deviation (SD) band to illustrate the dispersion at each time point of the gait cycle. A low SD indicates that the data points are likely to be very close to the mean. For normally distributed data, one SD on either side of the mean accounts for 68% of the data set. The published gait normative database which includes kinematics, kinetics and power was presented in this way, as the changes in the means and SD over a gait cycle were of interest (Kadaba et al., 1989; Perry, 1992; Schwartz, Trost, & Wervey, 2004; D. A. Winter & Yack, 1987; D.A. Winter, 2009).

7.4 Results

7.4.1 Speed dependence of the lower leg muscles

Eight healthy younger participants (a subset of 11 participants described in Chapter 5 section 5.3) whose walking data at five speeds were captured successfully were included in the analysis of the different speeds: five male and three females, with an average age 34±5 years, an average mass of 71±14 kg and averaged height of 170± 8 cm, were recruited. Their average walking speed was 1.18±0.15 m/s. The detected average speeds: slowest (0.64 ±0.15 m/s), slower (0.94 ±0.14 m/s), self-selected speed (1.18 ±0.15 m/s), faster (1.46 ±0.19 m/s) and fastest (1.86 ±0.24 m/s). The SD increased during the faster and fastest speeds. The SDs of the normal and slower speeds were similar. The timing of the toe-off occurred at slightly different percentages of the gait cycle depending on the speed of walking. The toe-off was at 62±2% of the gait cycle for the slowest walking speed, 61±2% for the slower speed, 60 ±1% for the self-selected speed, 61±3% for the faster speed and 58±3% for the fastest speed.

a) Characteristics of the normative EMG profiles of the lower leg muscles and other gait data at the self-selected speed

At the self-selected speed, the tibialis posterior showed biphasic activity during stance: minor activity approximately occurred at 10% of the gait cycle and the larger peak activity occurred at approximately 40%. The inter-subject SD during early stance activity (the first burst of activity) was higher than that during the second major peak (Figure 7.2). The average SD was 10% of the peak at normal speed. In the tibialis anterior, two bursts of activity were observed during the transition from swing to stance and during the swing phase, with a relatively high SD during the latter. Regarding the medial gastrocnemius, a single active period was displayed. The peak activity occurred at approximately 40% of gait cycle. The SD was large during an increase in activity but low after reaching the peak.

The EMG profiles of the tibialis posterior consisted of continuous activity after the initial contact while there was slight decrease in the external eversion moment. On initial contact, the foot was slightly inverted before returning to almost neutral during mid-stance. The tibialis posterior activity reached a peak at the same time as the medial gastrocnemius, at approximately 40% of the gait cycle, when there was the maximum dorsiflexion and external dorsiflexion moment, the medial and lateral displacement was less than 5º (the foot was almost flat on the floor) and there was slight internal rotation. The kinetic graphs (Figure 7.2) showed a reduction in the external eversion moment and internal rotation moment over this period. There was power absorption before the large power generated in the late stance.

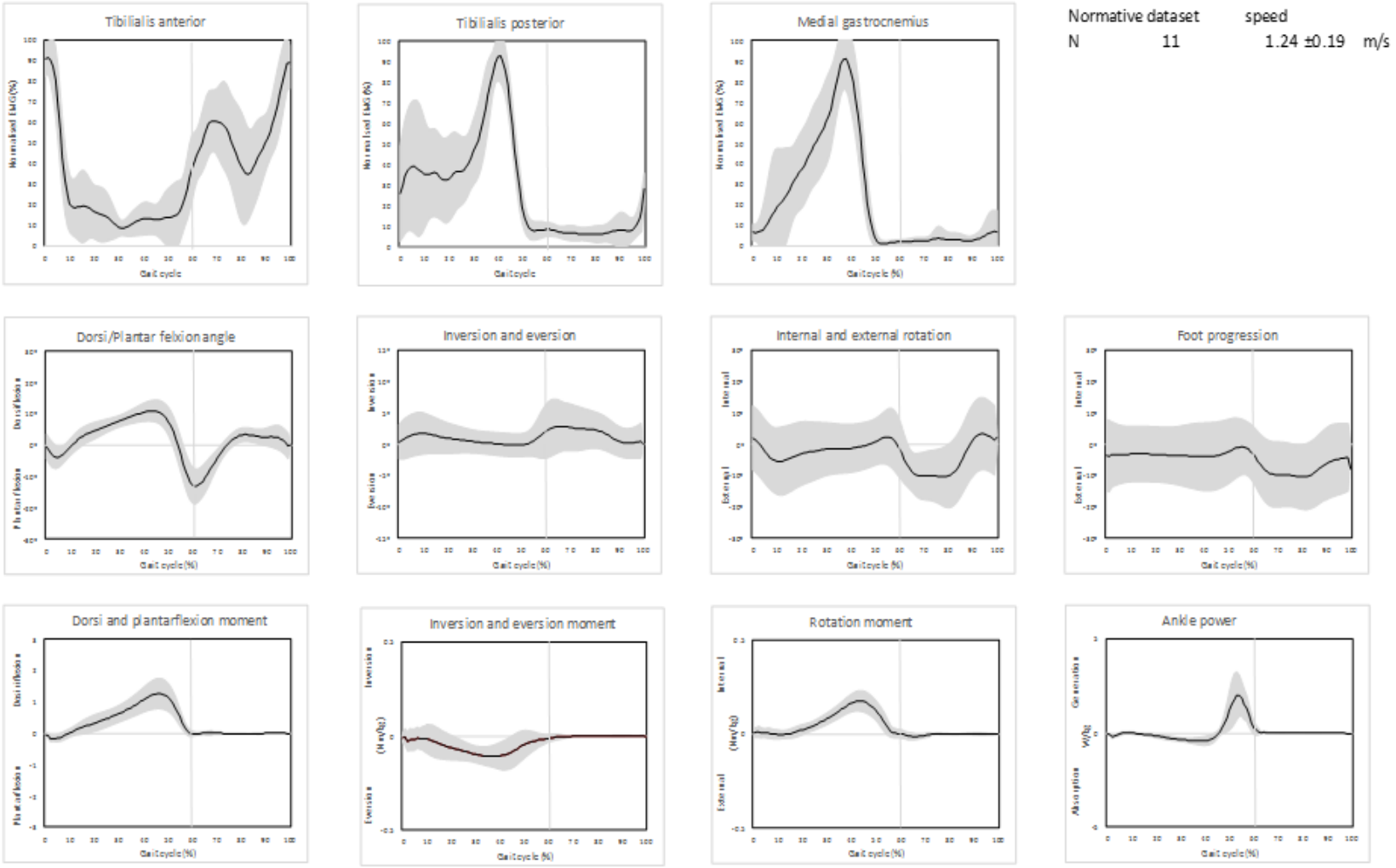


Figure 7.2 A normative dataset from 11 healthy participants at self-selected speed

b) Characteristics of the normative EMG profiles of the lower leg muscles across different speeds

The averaged correlation coefficient (r) between the self-selected speed and other speeds from all younger participants are from 0.819 at the fastest speed to 0.918 at the faster speed (Table 7.1). The EMG at the faster speed (a 25% increase) had a higher correlation to the self-selected speed than the other speeds. The tibialis posterior showed the most noticeable increase in peak amplitude at the fastest speeds (Figure 7.3-4). The timing of the peaks occurred earlier when walking at faster speeds: at 36% of the gait cycle for the fastest speed and at 42% of the gait cycle for the slowest and toe-off occurred earlier: at 58% of the gait cycle for the fastest speed and at 62% of the gait cycle for the slowest. When walking at faster speeds than the self-selected speed, the activity of the tibialis posterior during early stance became more prominent, particularly at the fastest speed. When walking at the slowest speeds (50% slower than normal), the activity of the tibialis posterior showed an almost single fluctuating burst of activity during stance. The SD across the gait cycle increased when the speeds altered. In particular, the SD at the fastest speed was maximal (24% of the peak recorded at the normal speed). The smallest SD was found when walking at the self-selected speed (Table 7.2). Between the slower speeds, the SDs were almost equal.

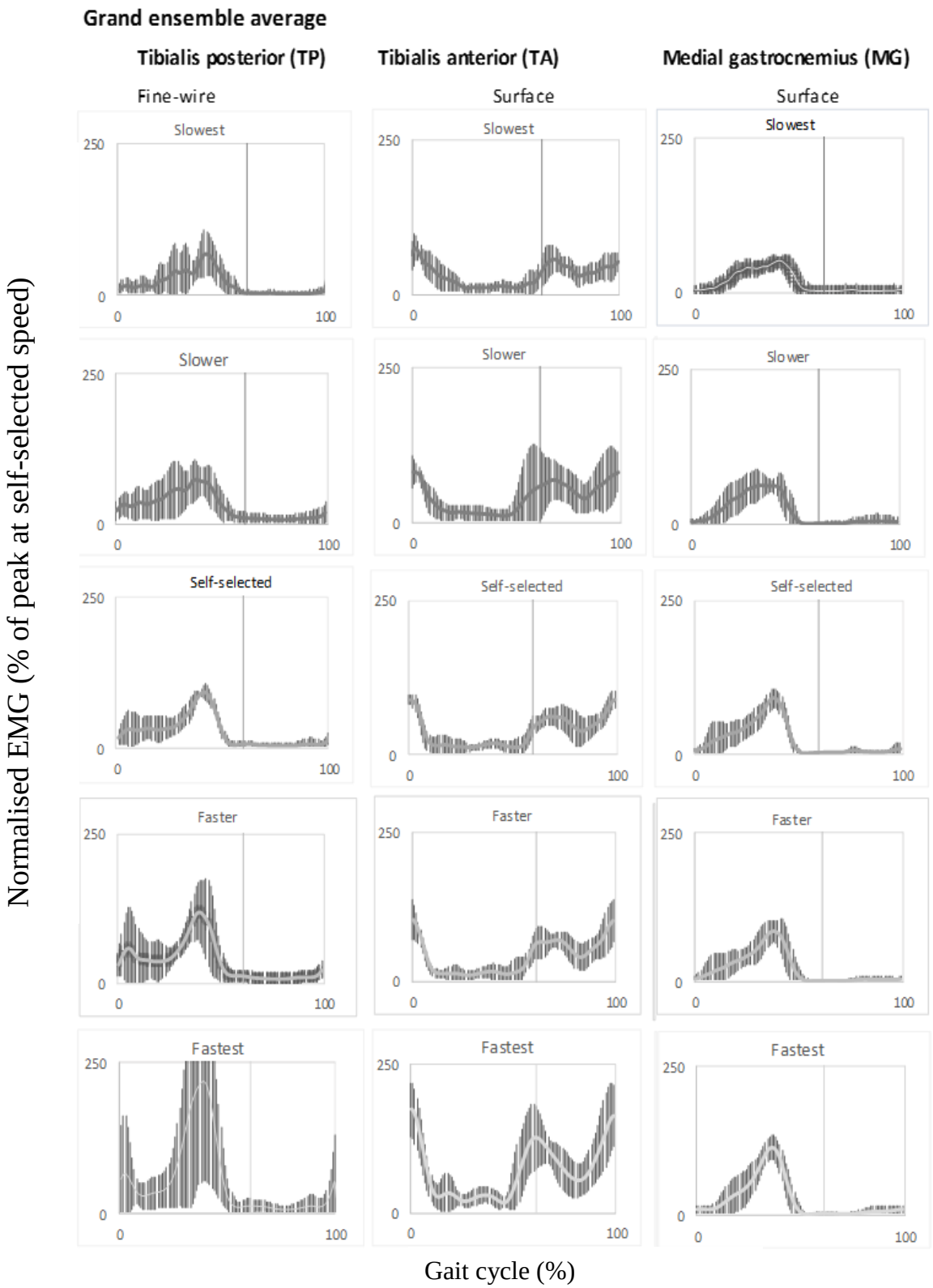


Figure 7.3 Grand ensemble average of the EMG profiles across five speeds

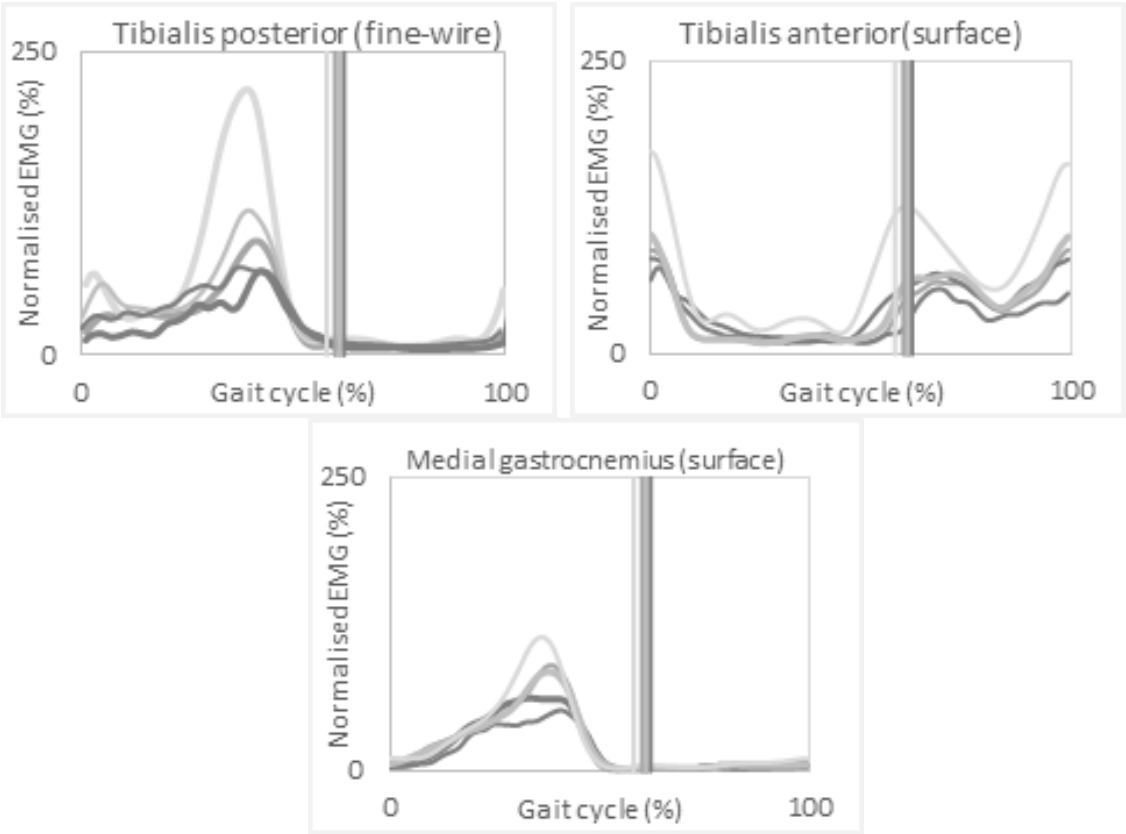


Figure 7.4 Grand ensemble average of the EMG at five speeds (the darkest line is from the slowest speed and the lightest line is from the fastest speed)

Table 7.1 Correlation coefficient for all speeds compared with normal speed

ID	*Tibialis posterior (fine-wire)*				*Tibialis anterior (surface)*				*Medial gastrocnemius (surface)*			
	Slowest	Slower	Faster	Fastest	Slowest	Slower	Faster	Fastest	Slowest	Slower	Faster	Fastest
1	0.690	0.755	0.847	0.952	0.610	0.714	0.795	0.783	0.593	0.653	0.823	0.947
2	0.749	0.544	0.969	0.544	0.500	0.357	0.882	0.357	0.869	0.755	0.930	0.755
4	0.822	0.968	0.969	0.912	0.865	0.862	0.925	0.814	0.835	0.989	0.943	0.964
6	0.798	0.944	0.900	0.780	0.870	0.974	0.964	0.708	0.943	0.990	0.983	0.983
7	0.929	0.905	0.904	0.917	0.833	0.960	0.916	0.877	0.984	0.963	0.976	0.990
8	0.948	0.950	0.914	0.913	0.856	0.946	0.927	0.892	0.979	0.972	0.961	0.878
9	0.713	0.871	0.928	0.665	0.818	0.898	0.972	0.934	0.656	0.716	0.967	0.647
11	0.960	0.966	0.914	0.867	0.960	0.978	0.960	0.845	0.967	0.956	0.897	0.845
Average	**0.826**	**0.863**	**0.918**	**0.819**	**0.789**	**0.836**	**0.918**	**0.776**	**0.853**	**0.874**	**0.935**	**0.876**

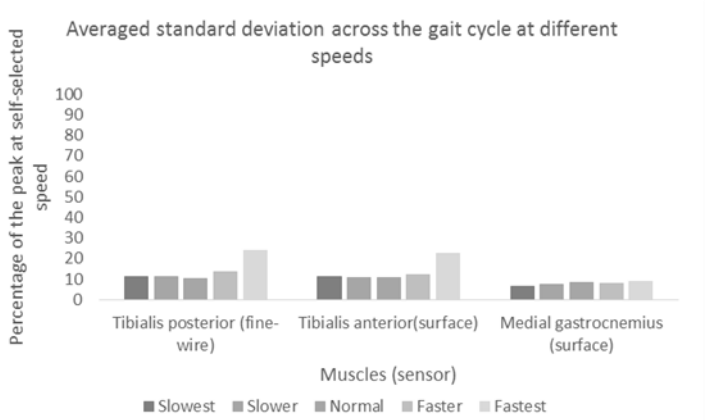


Figure 7.5 Averaged standard deviation across the gait cycle at different speeds

Generally, the peak amplitude tended to increase and the timing of the peak tended to occur earlier with faster speeds. However, the changes between speeds were not directly proportional. At faster and fastest speeds, there was an increase in amplitude and earlier timing of the peak recorded from three muscles, and the changes in the tibialis anterior and tibialis posterior were clearly noticed. Similar to the tibialis posterior, substantial increases in peak amplitude were found in the tibialis anterior and medial gastrocnemius at the fastest speed compared with the changes at other speeds. The amplitude change in medial gastrocnemius activity over a range of speeds was less than for the other muscles. At slower and slowest speeds, the amplitudes were slightly lower than those at the self-selected speed. In figure 7.1 and 7.2, the temporal patterns of the EMG profiles for the tibialis anterior and medial gastrocnemius across different speeds were approximately consistent. The averaged 'r' relative to the normal speeds range from 0.776 at the fastest speed to 0.918 at the faster speed for the tibialis anterior and range from 0.853 at the slowest speed to 0.935 at the faster speed for the medial gastrocnemius (Table 7.2). The EMG at the faster speed (25% increase) had a higher correlation to the self-selected speed than the other speeds.

In two out of three muscles, the SD was minimal during the self-selected speed and maximal at the fastest speed, with the medial gastrocnemius as an exception (Figure 7.1 and 7.3). The SD of tibialis posterior was the most sensitive to the speeds compared with the other

muscles. Similar to the tibialis posterior, the SD of the tibialis anterior grand ensemble averaged EMG profile was highest at the fastest speeds, then reduced at the slower, faster, slowest speeds accordingly and, at the normal speed, the SD was the lowest (approximately 10% of the peak). In the medial gastrocnemius, the SD was almost constant (approximately 10% of the peak measured at the normal speed).

There seems to be considerably greater variability in the speed in the EMG graphs than in the joint ankle and moment graphs. It is only really in regard to joint power where similar levels of variability are seen. The ankle angles, moments and power increased in range as the speed increased and the peaks also occurred earlier in the gait (Figure 7.6-11). The SD of the ankle angles over the range of speeds was approximately constant, though it increased slightly at faster speeds (approximately 1 degree and 0.1 Nm/kg) (Figure 7.10). On the other hand, the average SDs across the gait of the kinetic graphs were directly proportional to the speeds. Across different speeds, the patterns of movements were slightly affected compared with the change in the amplitude and timing. The changes of moment across the speeds were less clear than the power graph, which showed systematic changes. The profiles of the power and the SD of the ankle moments were the most sensitive to systematic speed changes.

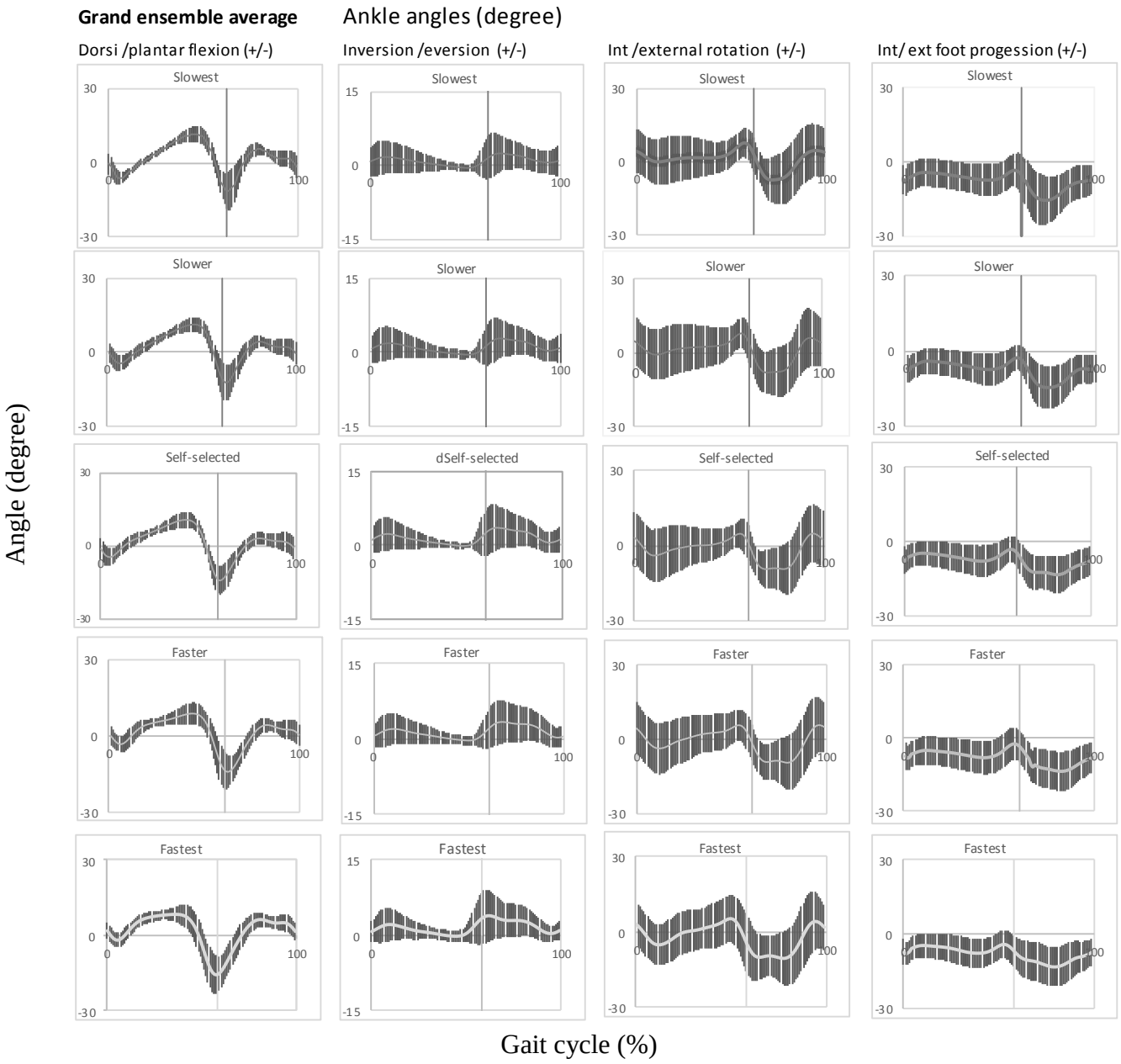


Figure 7.6 Grand ensemble average of ankle kinematics with standard deviation

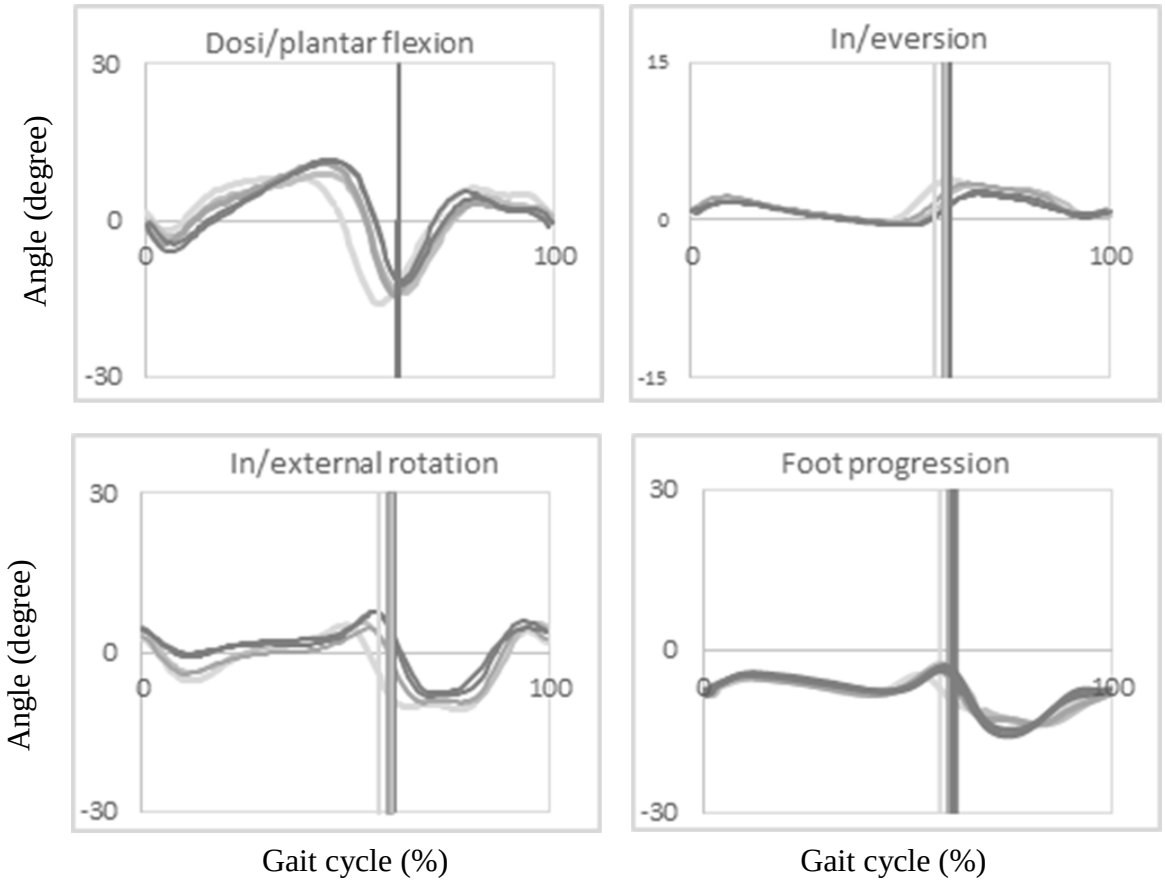


Figure 7.7 Grand ensemble average of ankle kinematics at five speeds

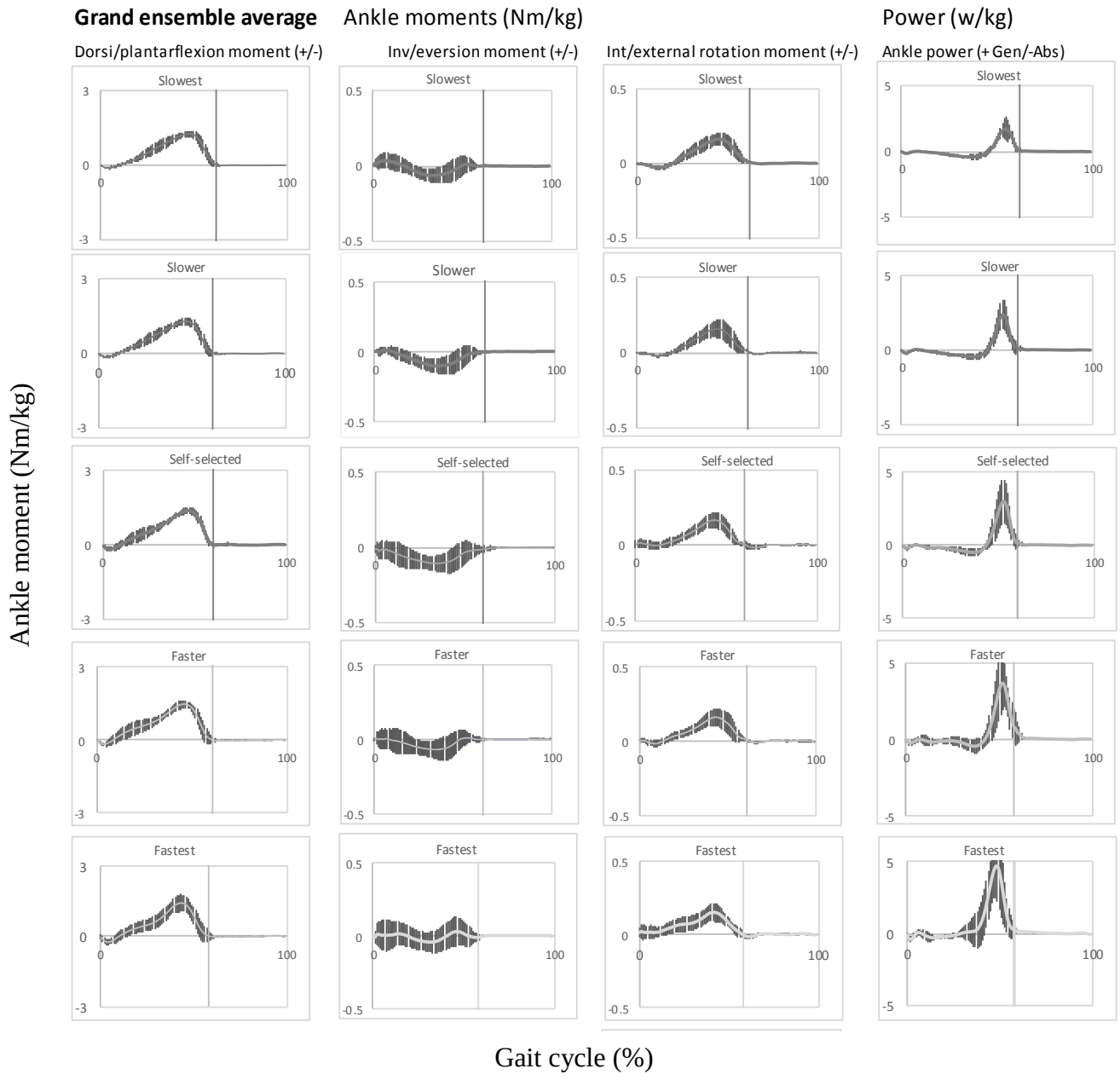


Figure 7.8 Grand ensemble average of ankle kinetics with standard deviation (the x-axis is the percentage of gait cycle and the y-axis is the moment in Nm/kg and the power in W/kg)

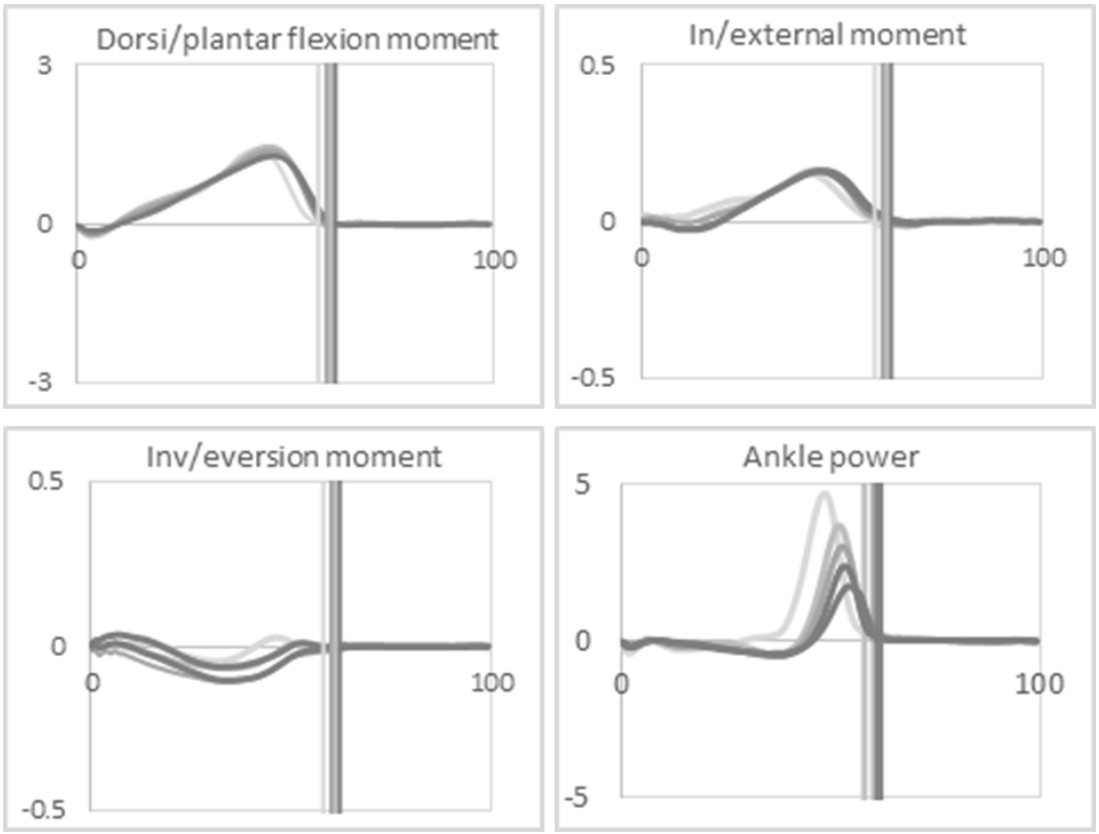


Figure 7.9 Grand ensemble average of the ankle kinetics at five speeds (the darkest line is from the slowest speed and the lightest line is from the fastest speed) (the x-axis is the percentage of the gait cycle and the y-axis is the moment in Nm/kg)

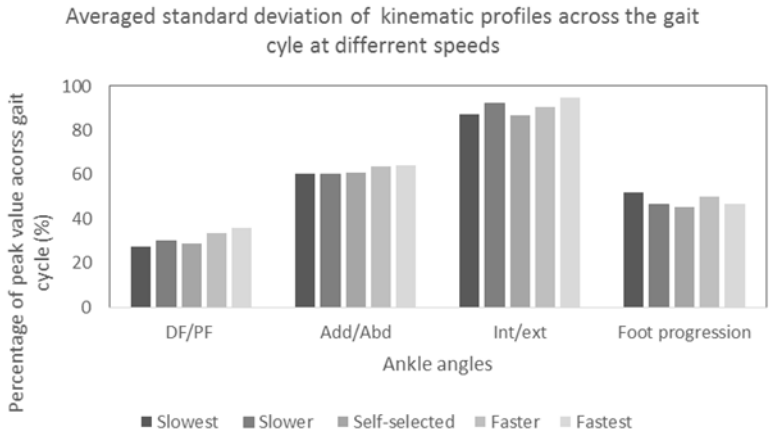


Figure 7.10 Averaged standard deviation of the kinematic profiles across the gait cycle at different speeds (DF/PF-dorsiflexion and plantarflexion angles, Add/Abd-adduction and abduction angles, and Int/ext-internal and external rotation)

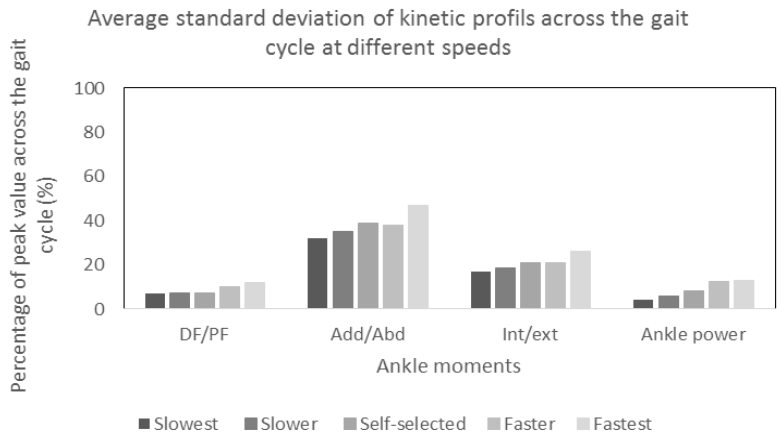


Figure 7.11 Average standard deviation of the kinetic profiles across the gait cycle at different speeds (DF/PF-dorsiflexion and plantarflexion angles, Add/Abd-adduction and abduction angles, and Int/ext-internal and external rotation)

7.4.2 Comparison of EMG data from older and younger adults

Four older healthy participants aged over 54 years old (one male, three female, averaged age 62±6 years with averaged mass 62±14 kg and averaged height 158±10 cm were recruited as a convenience sample. Their averaged walking speed was 1.03±0.21 m/s.

Figure 7.12 showed that EMG profiles and simultaneous ankle kinematics and kinetics from four participants with the age over 54 years were similar to those of younger normative dataset except activity of tibialis anterior and the inversion-eversion moment. The averaged inversion and eversion moment were calculated from three participants only. This was because the calibration wand for Healthy 02 was slightly deviated from the force plate border causing small degree of error from the origin and excessive inversion moment.

The patterns of EMG signals detected from the tibialis posterior and medial gastrocnemius were consistent with the normative dataset except for the profile of the tibialis anterior (Figure 7.12). At between 10% and 40% of the gait cycle, the tibialis anterior EMG was outside the younger SD band, at approximately 40% of peak amplitude, and there was a slightly early peak of activity during swing, at approximately 5-10%. The normalised EMG amplitude of the medial gastrocnemius and its variation was consistent with the normative

data and consistent peak amplitude between gait cycles within subjects. However, the variations in the EMG between subjects were relatively small. There was a slightly delayed peak of the tibialis posterior and slight variation in the activity of the tibialis anterior.

Regarding the other gait data, the ankle kinematics and most ankle kinetics were consistent within the normative dataset with small inter-subject variation, indicated by SD. There was a limited range of movement in the frontal plane and a higher inversion moment compared to the normative dataset. The graph showed the continuous inversion moment.

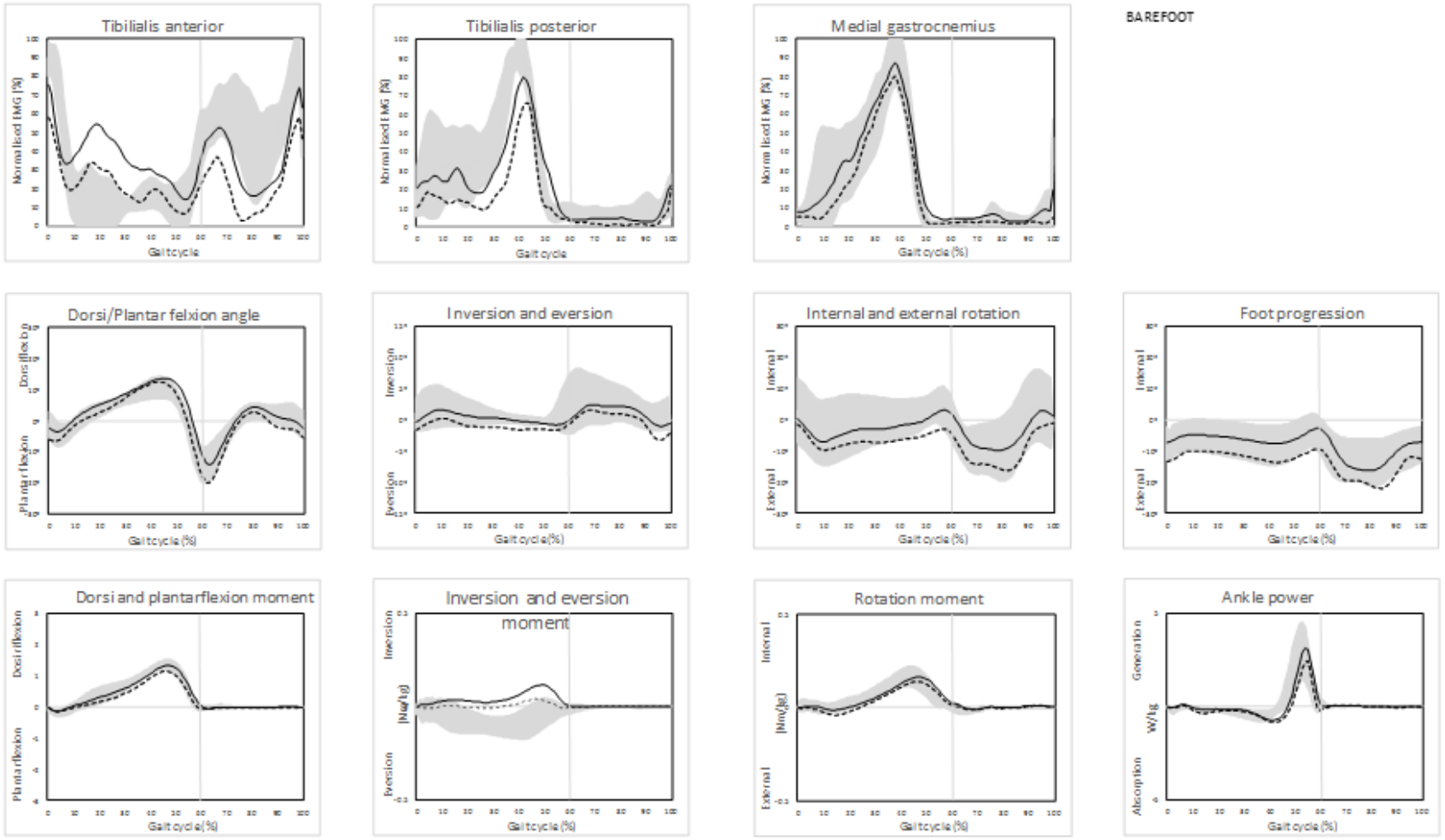


Figure 7.12 Comparison between the averaged data of the four older participants against the younger normative dataset
(The line is the grand ensemble average and the dash is the SD)

7.5 Discussion

7.5.1 Speed dependence of the lower leg muscles

a) Characteristics of the normative EMG profiles of the lower leg muscles and other gait data at self–selected speed

EMG profiles

The tibialis posterior acts as an invertor to maintain the medial longitudinal arch due to its insertions on the tarsal bones (Kaye & Jahss, 1991; Semple et al., 2009) and assists plantar flexion during gait. Our finding is in agreement with the recent studies that the tibialis posterior was active at 10% and 40% of the gait cycle, which was at the same time as the peak activity of the medial gastrocnemius, although there was higher within session, between-subject variability in tibialis posterior activation among healthy adults compared with the tibialis anterior and medial gastrocnemius (Murley, Buldt, et al., 2009; Ringleb et al., 2007). However, the amplitude of the first burst of tibialis posterior activity in our study was less than that reported in Murley's studies (Figure 3.2-3 in Chapter 3) but similar to the control group in Ringleb et al. (2007), which reported only stance phase data. The different normalised amplitudes between our data and Murley's data may be the result of foot posture and normalisation technique (MVIC) (Murley, Menz, et al., 2009). Our aim is to detect the pathological features within normal healthy variations but not specifically deviations from normal foot posture. Also, our protocol employed the peak normalisation technique instead of the maximal voluntary contraction, which was previously reported as being a difficult technique for patients (Barn, Turner, Rafferty, Sturrock, & Woodburn, 2013) and less reliable than the peak technique which could allow the identification of the pathologic features from the normal variation of healthy subjects rather than the absence of a normal arch (Chapter 5). The EMG pattern and low variability of the tibialis posterior (average within session, between subject SD 12%) in our study gives confidence that our protocol

can provide an EMG dataset for asymptomatic healthy adults with no restriction on different foot postures compared to the previous reports (average SD 30%) in young healthy adults (Chapter 3).

The pattern of the tibialis anterior profiles consisted of two bursts of activity (Figure 7.2). This pattern is consistent with the profiles reported by the existing literature included in the systematic review (Chapter 3). Using our EMG acquisition protocol resulted in lower average between-subject variability (SD) in tibialis anterior activation across the gait cycle at self-selected speed (14%)(Figure 7.2) while the average SD of the tibialis anterior reported in the systematic review was 24%.

The medial gastrocnemius as part of the tricep surae showed a single burst of activity to provide support for and propulsion of the swinging limb (Figure 7.2). This pattern was consistent with the profiles reported in the systematic review (Chapter 3). Using our EMG acquisition protocol resulted in lower average between-subject variability (SD) across the gait cycle at self-selected speed (10%), while the average SD reported in the systematic review was 25%. The lower SD in these profiles is likely to aid the detection of pathological features, given that the capacity to distinguish between normal and abnormal is one of the key requirements of any biomechanical measure (Baker, 2006).

EMG, kinematics, kinetics and power

The foot is regarded as a rigid segment in this model. Due to the proximal placement of the forefoot markers on the base of the second metatarsal in this experiment, the frontal plane movement is the movement of the midfoot and rearfoot. The forefoot movement has not been tracked (one of the limitations).With the activities of the other muscles as well, the frontal plane and transverse plane foot movement observed in the kinematic measurements of this study are consistent with those reported for the control group (age over 50 years) in previous studies (Ringleb et al., 2007; Woodburn, Helliwell, & Barker, 2002).

The timing of EMG activation in the tibialis posterior compared to the joint moment profiles indicated that the former mainly contributes to moments in the frontal plane (inversion and eversion) and the tibialis anterior mainly to sagittal plane moment (dorsiflexion and plantarflexion) and inversion (Figure 7.2). The control of this external moment after initial contact is likely to be part of the foot loading mechanism (Barn et al., 2013; Perry, 1992). The activity during mid-stance was probably a mechanism to prevent the collapse of the medial longitudinal arch (Keenan, Peabody, Gronley, & Perry, 1991). Perry et al. (1992) suggested that it is a dynamic locking of the subtalar joint for the provision of forefoot support. Also, the tibialis posterior as part of the plantar flexors may contribute to these plantarflexor moments alongside the medial gastrocnemius and others to provide push off. Different participants may have different activation patterns of invertors and evertors (e.g. the peroneus longus) and muscle co-contraction to control this external moment, which will be greater or lesser in healthy individuals according to normal variations in foot, ankle and knee alignment, for example.

b) Characteristics of normative EMG profiles of the lower leg muscles across different speeds

Walking at faster speeds tended to result in a higher demand for ankle power generation, higher muscle activity to control higher external moments and less time being available to incorporate peripheral feedback (Figure 7.3-9). The stance phase tended to be shorter when the speed increased, so the timing of peak of these gait parameters was likely to occur earlier in the gait cycle. The most noticeable changes in our dataset were seen in the tibialis posterior amplitudes and generated ankle power while there was slight increase in medial gastrocnemius activity and small change in the ankle moments. However, the differences between the speeds in our data were less systematic than those previously reported. This may be because earlier researchers' methods of speed controls are more precise through the

use of post-hoc techniques to categorise each trial based dimensionless speed with standard deviation (Schwartz et al., 2008) and a treadmill (Stoquart et al., 2008; Van Hedel et al., 2006).

Similar to previous studies of EMG in the tibialis posterior and other muscles, the amplitude of the EMG signals tended to increase when the walking speed increased (Den Otter et al., 2004; Murley et al., 2014; Schwartz et al., 2008; Warren et al., 2004), probably due to the higher muscular demand (moment) to control medio-lateral foot stability (a small increase in the inversion and eversion angles) and propulsion. In our study, the amplitude of the second burst was directly proportional to the speed and considerably increased at the fastest speed. This was similar to changes in ankle power. Therefore, the activation of the tibialis posterior contributed to ankle power generation-propulsion.

The timing of the peaks occurred earlier when walking at faster speeds as the percentage of the stance phase was shorter when walking at faster speeds, similar to other muscles (Figure 7.3-4). This finding is different from previous reports of the tibialis posterior across different speeds. They found a significantly delayed timing of the peak of tibialis posterior activation when walking at faster speeds, which was different for the other muscles reported in the same study (Murley et al., 2014) (Figure 7.1). Our speed range (0.64-1.86m/s) is slightly wider than theirs (0.67-1.63 m/s), so the difference between the data is unlikely to be due to speed. The delayed timing as the speed increases may be due to a specific neural foot posture in Murley's studies. Therefore, the EMG profiles of the tibialis posterior for healthy adults during gait have similar speed-related responses as the other muscles in the lower limbs: an increase in amplitude and early timing of the peak when the speed increases.

In the tibialis anterior and medial gastrocnemius, the amplitude of the EMG tended to increase and the timing of the peak occurs earlier when walking at faster speeds than normal (Figure 7.3-4). This is in agreement with other studies e.g. Den Otter et al. (2004); Hof et al.

(2002); Shiavi et al. (1987a). The increase in the tibialis anterior at faster speeds is probably due to the higher demand for the dorsiflexing moment of the foot over a shorter period of time. The changes in the amplitude and timing of the peaks were somewhat linked to speed in our study but, at the fastest speed, these changes were considerably higher than for the other speeds, regardless of the systematic speed alterations. This may represent the upper limit of walking speed before the alteration of the muscle co-activation patterns occurs to support different modes of locomotion (e.g. jogging). Hof et al. provided the option to predict EMG profiles across speeds for 14 muscles, including the tibialis anterior and medial gastrocnemius (Hof et al., 2002). The slight inconsistency in the literature regarding the systematic relationships between EMG and walking speed may be due to the different ways of controlling the speed (e.g. a metronome, practising walking a certain distance within a given time).

In the medial gastrocnemius, our study and the previous studies agreed upon the major burst of activity during mid-stance of which the amplitude was proportional to the walking speed (Den Otter et al., 2004; Murley et al., 2014; Warren et al., 2004). However the increase in amplitude was less than that for the other muscles in the same studies. This was also found in (Murley et al., 2014; Nymark et al., 2005; Shiavi et al., 1987a; Warren et al., 2004). This was probably because the medial gastrocnemius was one of several plantarflexor muscles (with the lateral gastrocnemius, soleus and tibialis posterior) contributing to forward propulsion and at the slower and slowest speeds, when the forward momentum was low during the propulsion phase, a similar amplitude of the medial gastrocnemius was required with a small adjustment of the smaller muscles.

The within session, between subject SD in EMG may be due to the different levels of activity required to walk at targeted speeds or different kinetics between individuals (Yang & Winter, 1985). Similar to our study, other studies also found non-linear relationships

between the variability of EMG and walking speed (Van Hedel et al., 2006). In our study, the lowest SD calculated from the EMG profiles was measured at a self-selected speed. One of the reasons for the least variability at the self-selected walking speed may be because the included participants tend to utilise similar motor programming/muscle coactivation patterns for the tibialis anterior and tibialis posterior with optimised energy cost (Mcneill Alexander, 2002; Waters & Mulroy, 1999). The dataset collected at the faster or slower speeds would be less sensitive to pathological changes than the dataset at the self-selected speed due to the higher SD. However, if the EMG difference is detected between our speed-matched dataset and the patient walking at self-selected speed/optimised energy cost and the control, the difference is substantial.

The overarching aim of this thesis was to provide protocols to collect a normative dataset for the clinical application of fine-wire and surface EMG with kinematics and kinetics. Our data and the previous reports (Kadaba et al., 1985) agree that EMG should be collected from a patient walking at a self-selected speed in order to provide reliable data (lowest variability). Then, the patient's EMG should be compared to a speed-matched dataset, in which the amplitude and timing of the peak were altered. This is to ensure that the detected difference is not because of the different speeds. For CGA laboratories, the EMG protocols should be followed in order to allow comparisons between subjects, sessions and laboratories (chapter 5 and 6).

7.5.2 Comparison of EMG data from older and younger adults

In older participants: the self-selected speeds were slightly slower than the average speed of younger adults but were still within the range of self-selected speeds seen in our younger adult participants, similar to the previous report (Schmitz et al., 2009). This may be because our participants were active and healthy, without any underlying pathology. The significant

changes in speed were seen in adults over 65 years old (Kerrigan, Todd, Della Croce, Lipsitz, & Collins, 1998; Mcgibbon & Krebs, 1999) so any differences seen in muscle activity in our data are unlikely to be caused by the different speeds of younger and older adults, respectively.

The between-subject variability of the grand ensemble EMG profiles in all three muscles was low, indicating consistent patterns between these older individuals in this study (Figure 7.5). Although the number of participants was small, the grand ensemble EMG profiles are somewhat promising in helping to decide whether we need the EMG profiles for older adults in the dataset.

Despite the age-related changes to the neuromuscular system, we found that the patterns of the medial gastrocnemius and tibialis posterior from older adults were within the range of younger adults but that the tibialis anterior profiles showed definitive differences in EMG profiles during specific parts of the gait cycle (Figure 7.12). This may be caused by the reduced strength of the tibialis anterior and the need for a greater number of recruited motor units to provide sufficient strength/moment. Generally, the remodelling of the motor system with age is likely to decrease the overall amplitude of the EMG signal. However, this absolute amplitude changes due to aging would not be seen in our study because of the peak normalisation technique which is shown to result in comparable EMG profiles between healthy subjects and between sessions (chapter 5). The normalised EMG by peak amplitude effectively showed patterns which were consistent with the raw EMG and relative amplitudes of different bursts of activity (chapter 5). Based on our EMG protocols using peak normalisation, it was unnecessary to provide an age-matched dataset for the tibialis posterior and medal gastrocnemius, as significant differences were not found.

In the tibialis posterior profile, the slightly delayed peak (but still within the range of the younger dataset) may relate to the increase in inversion moment for foot stability (the

position of the foot was within the younger range) as, in older adults, the foot arch may have collapsed due to ligament laxity (Moore, Dalley, & Agur, 2013) (Figure 7.12). The amplitude of the tibialis anterior EMG increased during 10-40% of the gait cycle compared to the younger adult profile. A similar result was found in Schmitz et al. (2009), who suggested that this may be the result of the higher co-activation of the soleus and tibialis anterior during mid-stance to stiffen the ankle due to balance concerns (Benjuya, Melzer, & Kaplanski, 2004; Hortobagyi & Devita, 2006).

The early timing of the tibialis anterior peak amplitude in swing in older compared to younger adults may be a result of alterations in the motor units, contractile properties and neural pathways seen in the natural ageing processes. The progressive reduction of the motor neuron numbers in older adults decreases the number of functional motor units and increases the number of abandoned muscle fibres (Tomlinson & Irving, 1977). These fibers are degenerated or re-innervated, probably by the agents released from the muscle fibres deprived of an innervation resulting in a higher innervation ratio, larger motor unit and twitch force (Enoka, 2008; Keller-Peck et al., 2001; Masakado et al., 1994). Moreover, there is evidence that the frequency of MUAPS in older adults decreases with a similar variation in young adults (Erim et al., 1999)). Motor unit synchronisation is unaffected by aging (Kamen & Roy, 2000). Connelly et al (1999) illustrated a slower MUAP frequency and longer twitch contraction in older adults in the tibialis anterior at different levels of voluntary contractions. Therefore, to accommodate this electromechanical delay, the muscles have to be activated sooner to generate the necessary force for foot clearance. Different muscles may have different responses to aging due to the different fibre compositions, functions in fine movement or postural controls.

A slightly reduced range of movement: plantarflexion and inversion was seen in older compared to younger adults and may reflect a strategy to increase stability (Nigg, Fisher, &

Ronsky, 1994) (Figure 7.12). An increased inversion moment may be a result of the laxity of the plantar ligaments, inducing the collapse of the foot arches (Moore et al., 2013) and more external rotation of the foot when the foot progression is still in the normal range. However, whether or not the differences between older and younger adults in the inversion moment is reliable is in question owing to the fact that these structure are complex (and so may not be fully captured by the foot marker models used here which assume rigid segments) and the differences in the moments between the groups is small.

7.5.3 Limitations

The tibialis posterior, tibialis anterior and medial gastrocnemius are acting on the ankle and foot complex. There are several joints within the foot, including 3 major articulations: the subtalar, mid-tarsal and metatarsal phalangeal joints. The Plug-in gait models may not be specifically designed to track all of these movements. However, this study aims to assist clinical gait analysis where the Plug-in gait model is one of the most common models used to record the lower limb movements and present adequate information for this experiment i.e. the foot relative to the tibia. The use of a multi-segment foot model may help to investigate the finer movements occurring within the rearfoot and forefoot. Regarding the comparison between younger and older participants, these older participants were healthy and active. The general population belonging to this group may be less active, so the difference may be more pronounced.

7.6 Conclusion

Regarding the research questions:

i) How does walking speed affect the activity of the muscle of lower leg and associated kinematics and kinetics?

Our data showed that EMG amplitude and range of kinematics and kinetics increase and the timing of peak activity occur earlier in the gait cycle based on the calculation from PIG model, when walking speed increases.

ii) Is there any evidence that the neuromuscular changes associated with aging are manifested in the EMG of the lower leg muscles during walking?

The difference between the younger and older participants could not be identified for the tibialis posterior and medial gastrocnemius. Our data showed the differences in tibialis anterior: excessive activity during early stance and the early timing of the peak during the swing phase within our active older participants.

Chapter 8 Demonstration of EMG measurements using fine-wire and surface sensors in lower limb muscles for clinical gait analysis

The previous chapters of this thesis defined the rigorous data capture, analysis and presentation protocols for the incorporation of both fine-wire and surface EMG measurements with kinematic and kinetic measurements for clinical gait analysis (CGA). EMG signals from the tibialis posterior, tibialis anterior, and medial gastrocnemius for young healthy adults at five speeds were then collected using these collection and analysis protocols to be used as a reference for CGA purposes. This chapter will present three case studies using these collection procedures and a normative healthy dataset to illustrate how they might be applied in the clinical gait analysis of stroke participants walking with and without ankle-foot orthoses at the same speed range.

8.1. Background

8.1.1 EMG acquisition protocol

The collection of EMG from muscles which aid the identification of pathological changes in CGA has the potential to be useful in the assessment of the severity or nature of impairments, progress monitoring and the prediction of outcome intervention (Baker, 2006). In order to establish whether EMG is helpful within CGA, the variability around the normative reference data should be sufficiently small to allow for the detection of pathological muscular activation as EMG profiles falling outside the normative variability at time points in the gait cycle which correspond to impairment, walking context or treatment effect. This chapter will test this by examining the EMG of patients with neuromuscular impairment (stroke), where the EMG would be expected to be affected. Further, it will

examine the extent to which EMG reflects (via differences with respect to normative reference data) specific impairments of a drop foot which mainly affect the swing phase of the gait cycle and walking with conventional treatment-AFOs which prevents a drop foot by limiting ankle movement (kinematics).

8.1.2 Stroke

The World Health Organisation defines a stroke as 'rapidly developing clinical signs of focal and at times global disturbance of cerebral function, lasting more than 24 hours or leading to death, with no apparent cause other than that of vascular origin' (Hatano, 1976). In the UK, the incidence is about 152,000 a year (Townsend et al., 2012). The major causes are ischaemic, which accounted for the majority (85%), and haemorrhagic (15%) (Feigin et al., 2013).

Stroke participants may be affected by speech deficit, depression, neuropsychological disorders, functional difficulties and mobility problems. Following stroke onset, heamorrhage or thrombus affecting the cerebral arteries cause damage to the upper motor neurones, usually on one side of the brain (Olney & Richards, 1996). This immediately affects the ability to generate the normal amplitude of the voluntary contraction and timing of muscle activity (Olney & Richards, 1996). In general, the muscle activity reflected by EMG signals on the affected side is lower than on the contralateral side (Tang & Rymer, 1981).

A continuous reduction of the motor units, possibly due to a lack of trophic inputs, was found from 9 days after stroke onset and continued for a year (Hara, Masakado, & Chino, 2004). Dattola et al. (1993) found evidence of the structural rearrangement of the motor units after denervation and re-innervation following stroke, which may result in a mismatch between muscle fibre type and motor neuron characteristics. They also found a high

proportion of slow twitch motor units in the lateral gastrocnemius. Moreover, the firing frequency of motor units reduces after stroke (Chou, Palmer, Binder-Macleod, & Knight, 2013; Hu, Tong, & Hung, 2006), probably owing to a higher level of recurrent supra-spinal inhibition. Spasticity can also develop over a period of weeks (Dietz & Berger, 1984).

Similar to the aging process, the atrophy of type II fibres has also been found in stroke participants without hypertrophy of type I (Hachisuka, Umezu, & Ogata, 1997). With muscle atrophy, an increase in the number of units will be required to achieve a given force (Garland, Gray, & Knorr, 2009) and patients may perceive higher effort. Also, the loss of fast twitch motor units and the type II muscle fibres may result in a slow speed of contraction-slow movement and muscle weakness (Garland, 2008). Moreover, there is a notion that high co-contraction (the concurrent activation of agonist and antagonist muscles) can reduce the force generated by the agonist muscle partially because there is impairment in the reciprocal inhibition modulated by the supra-spinal control. These motor impairments could result in slow, weak force production and insufficient control of the functional task movements. Defective muscle activation, due to stroke, which results in weakness in voluntary contraction, is defined as *muscle paresis* (Lamontagne, Malouin, Richards, & Dumas, 2002).

All of these changes have been seen to affect coordination in terms of the modulation of agonist and antagonist muscle activities, and hence affect the quality of movement (Garland, 2009). Therefore, EMG on both legs could be affected by these changes and be manifested as decreased amplitude at higher frequency and changes in the timing of the activation of individual muscles.

8.1.3 Neuromuscular activation changes after stroke

Many studies report the abnormality of EMG profiles on both affected and sound limbs following stroke e.g. Buurke et al. (2008); Den, Geurts, Mulder, and Duysens (2007); Shiavi, Bugle, and Limbird (1987b), with a range of different pathological changes being identified e.g. Burridge, Wood, Taylor, and Mclellan (2001); Perry, Waters, and Perrin (1978). Commonly identified pathological changes are (1) the absence or reduction of EMG amplitude at the certain period of gait cycle: e.g. the absence or decrease of tibialis anterior activity during the transition from swing to stance (Burridge et al., 2001; Perry, 1993) and the short active period of the tibialis anterior during single support (Den et al., 2007), (2) the prolonged activity of muscles: e.g. the prolonged activity of the medial gastrocnemius during the double support on the paretic side (Den et al., 2007), and (3) the early timing of activity, e.g. early timing of plantarflexors (Perry et al., 1978). Only one study showed the activity of the tibialis posterior in participants with stroke and only then as on-off activity (Perry et al., 1978). They found that the activity was absent in almost half of their participants and, when present, it was active in the late swing and continued in the stance phase. This is in comparison to a complete absence in the tibialis posterior during the swing phase in healthy participants. This paper showed that tibialis posterior activity was affected but varied between patients (Perry et al., 1978).

8.1.4 Kinematic, kinetic and power changes to gait after stroke

The variations in kinematic and kinetic gait deviations resulting from altered neuro-muscular activation depend on the stroke severity, the location of the lesion, the time since the stroke, the prescribed types of rehabilitation, and other individual differences (Olney & Richards, 1996). Overall, the speed is reduced to between 0.18 and 0.74 m/s in 11 studies reviewed by Tyson and Kent (2013). Common gait deviations in stroke participants included the

reduction of the sagittal dorsiflexion/plantarflexion range, excessive knee flexion (e.g. weak knee extensor) or knee hyperextension (e.g. spastic plantarflexors or adaptation for stability) in the stance phase, insufficient knee flexion in the swing inducing other gait deviations to assist ground clearance such as hip hiking and circumduction (Olney & Richards, 1996).

Twenty percent of stroke participants were estimated to have *a drop foot*, a common impairment characterised by the failure to dorsiflex the ankle due to weakness and altered neuromuscular activation patterns of the dorsiflexor tibialis anterior (Burridge, Taylor, Hagan, Wood, & Swain, 1997; Kottink et al., 2004). Drop foot may present as a spastic equinovarus foot which is caused by shortening and over activity of the plantar flexor muscles (e.g. gastrocnemius or soleus), foot invertors (tibialis posterior and tibialis anterior) and toe flexors (flexor digitorum longus/brevis and flexor digitorum longus/ brevis) (Ward, 2014). This commonly results in mid-foot or fore-foot contact and poor push off (Wong et al., 2004). Stroke patients with a drop foot may experience instability in the stance phase, thereby making it difficult to provide stable support for forward progression, difficulty in foot clearance which may cause tripping in the swing phase and difficulty in avoiding obstacles (Chen, Patten, Kothari, & Zajac, 2005; Ohata, Yasui, Tsuboyama, & Ichihashi, 2011; Olney & Richards, 1996; Simpson & Jiang, 1999). There are many possible causes for this, such as weakness in the tibialis anterior and peroneus longus or over-activity of plantarflexors, tricep surae and tibialis posterior.

The impaired single limb support causes a shorter support time for the paretic limb with exaggerated propulsion by the non-paretic limb during pre-swing to minimise the swing duration. In the paretic limb, Chen et al. (2005) described impaired swing initiation resulting in prolonged swing time, insufficient knee flexion at toe-off and mid swing. This leads to gait deviation, such as hip hiking and circumduction for clearance. The spasticity of the

plantarflexors may further limit the ankle dorsiflexion with weak knee extensors and hip flexors (Dickstein, 2008; Stein et al., 2010). With insight into muscle activity, the EMG of the tibialis posterior, tibialis anterior and medial gastrocnemius, the contribution of each of these muscles to these kinematic and kinetic gait deviations can be determined or distinguished from other biomechanical causes. Thus, the addition of EMG to the standard kinematic and kinetic CGA practices could assist in the identification of the cause of impairment and targets for treatment, and also help to quantify the effects of the treatment.

The studies above demonstrate that patients with stroke can present with a wide range of kinematic and kinetic impairment. CGA based on these measures alone is often insufficient to indicate clearly the specific underlying impairments and therefore indicate the targeted treatment. Similarly, what EMG studies there have been also show wide variations in muscle activation patterns. It is thus clear that methods integrating kinematics and kinetics measurements with EMG using a standardised protocol for collection, analysis and interpretation is needed.

The purpose of this chapter is, firstly, to provide proof-of-concept that the normative EMG profiles gathered in previous chapters using standardised protocols are sufficiently sensitive to be used as diagnostic reference comparisons for patient data. This will be established using both data from a healthy older adult (for whom deviations in EMG beyond variability seen in healthy younger adult data is not expected) and a case series of stroke patients with known neuromuscular pathology (for whom deviations in EMG profiles are expected to be greater than the variability in healthy reference data). Secondly, to explore if EMG varies with kinematic and kinetic changes, two comparisons are made: (1) comparison of EMG profiles, kinematics and kinetics between normative profiles and stroke participants with gait deviation; (2) comparison of EMG profiles, kinematics and kinetics between stroke participants walking with and without AFO. This tests the ability to detect the changes due

to mechanical restriction (use of various degrees of control provided by AFOs to limit plantarflexion during swing and stabilise the frontal movement of the ankle-foot complex) and those due to neuromuscular impairment. Overall, these comparisons will help to explore if EMG data can help explain kinematic and kinetic gait deviations and provide proof-of-concept for the use of EMG in standard CGA practice.

8.2 Research questions

i) Is the variability around normative reference EMG collected using our suggested protocols sufficiently low to detect differences in the EMG profiles of the tibialis posterior, tibialis anterior and medial gastrocnemius due to neuromuscular pathology (stroke)?

ii) What does the EMG report add to an analysis of kinematics and kinetics in CGA?

- Can EMG profiles help to explain the gait deviations seen in kinematic and kinetic graphs between normative data and participants with post-stroke?

- Can EMG profiles show the difference between walking with and without AFO for participants with post-stroke?

8.3 Method

Data were collected using the protocols outlined in Chapter 4 and Chapter 5 for peak normalisation with the following exceptions.

8.3.1 Participants

One healthy participant aged 69 was recruited from the University of Salford, and three participants post-stroke were recruited from a convenience sample who attended the Brain and Spinal Injury Centre (BASIC) in Salford, UK. Inclusion criteria for the participants with stroke included: at least 6 months post-stroke, wearing a prescribed AFO and having the

ability to walk without other walking aids, as this may cause changes in EMG (Buurke, Hermens, Erren-Wolters, & Nene, 2005), stable medical status and the ability to give informed consent. Exclusion criteria for all participants included: participants who were younger than 18 years, the presence of other medical conditions: neurological, rheumatic, cardiovascular including diabetic mellitus, injury to the deep or superficial peroneal nerve and sciatic nerve, any medical conditions that affect the use of fine-wire electrodes and pregnancy. Ethical approval (HSCR14-100) was obtained from the University of Salford Research Ethics Committee. All participants provided informed consent prior to their participation.

8.3.2 Protocol

The participants were asked to bring their shoes and AFO (for stroke participants). The study used an instrumented gait lab for ankle angle, ankle moment, and EMG on the tibialis anterior and medial gastrocnemius using a surface electrode and the tibialis posterior using a bipolar fine-wire electrode on both limbs. All participants were asked to walk at a self-selected speed along a 6 m walkway in the Gait Laboratory at their self-selected speed. The participants with post-stroke were asked to walk with and without their AFO and combined footwear. Individuals were given approximately one minute rest after each trial and whenever required. Six successful walking trials, when the each foot entirely contacts different force plates, were collected when walking, with and without AFO. The individual ensemble average EMG profiles were normalised by their averaged peak values collected during barefoot walking to allow comparison between individuals and the normative dataset and also to investigate the effect of AFO on EMG.

8.3.3 Data analysis

The mean and standard deviation were plotted against the percentage of the gait cycle. After processing the EMG signals, they were presented as a mean or an ensemble average and standard deviation (SD) band to illustrate the dispersion or spread from the average. The grey bands in the graphs are the mean and ±SD from speed-matched definitive normative profiles from chapter 7. The comparison of ensemble averages and corresponding SD were made between: (1) healthy older participants and the speed-matched normative database, (2) case series of participants with post-stroke and the speed-matched normative database, and (3) walking with and without AFO in participants with stroke. A mark-up system from the impairment focused approach (Baker, 2013) was used to compare normal data from previous chapters and patient data when the mean fell outside the ±SD bands : ⊕, ○ too much/little for part of the cycle; ⬆ ,⬇ too much/little throughout the cycle; ➡ , ⬅ too late/early; ⬌, ➡⬅ too long/short; ◿, ◺ abnormal slope; ⬭ other; red for the paretic side, green for the sound side and blue for both sides.

8.4 Results

8.4.1 Case 1: Healthy older adults

The healthy older adult participant in the study was aged 69 years, with a mass of 55 kg, and height of 152 cm. Her self-selected walking speed was 1.0 m/s within the self-selected speed range of our young dataset. On first inspection, the gait graphs of the healthy older participant show activity of the tibialis posterior and medial gastrocnemius, and ankle kinematics and kinetics within the +/-1SD bands of younger healthy adults at matched-speed for the majority of the gait cycle. The activity of the tibialis anterior is above these bands from 10-30% of GC (Figure 8.1 marked a) and the dotted standard deviation bar for the participant's data suggests that this is consistent across walking trials.

This pattern of EMG activity in the tibialis anterior in the early stance, however, invites closer inspection of other graphs. Whilst the EMG signal for the gastrocnemius lies within the +/-1SD bands, it does show a plateau of activation of the 10% peak signal through the first double support and only begins to increase above this in early single support. As might be expected, this is mirrored in the ankle moment which shows a delay in the generation of the development of a plantarflexion moment through early single support. Close inspection of the ankle kinematics also shows that there is minimal plantarflexion in the first double support followed by early dorsiflexion which persists into early single support.

In summary, the data suggest that prolonged tibialis activation is associated with delayed activation of the gastrocnemius, resulting in a delay in the development of the plantarflexion moment. These changes appear to occur shortly after the changes in the ankle kinematics, suggesting that these might be an adaptation to the way in which the foot lands and the load is accepted onto it.

These data are from an asymptomatic participant with no diagnosed neuromusculoskeletal pathology and suggests a mechanism that is within the normal range of variability. It is important to recognise that the +/-1SD limits, although universally plotted within clinical gait analysis, only represent 65% of the normative range and that such features within gait data from healthy participants are to be expected. In the context of this thesis, however, it is important to note that the comparatively mild kinematic and kinetic features in early stance would probably not have been commented upon without the larger features in the EMG data, particularly that for the tibialis anterior, to draw attention to them.

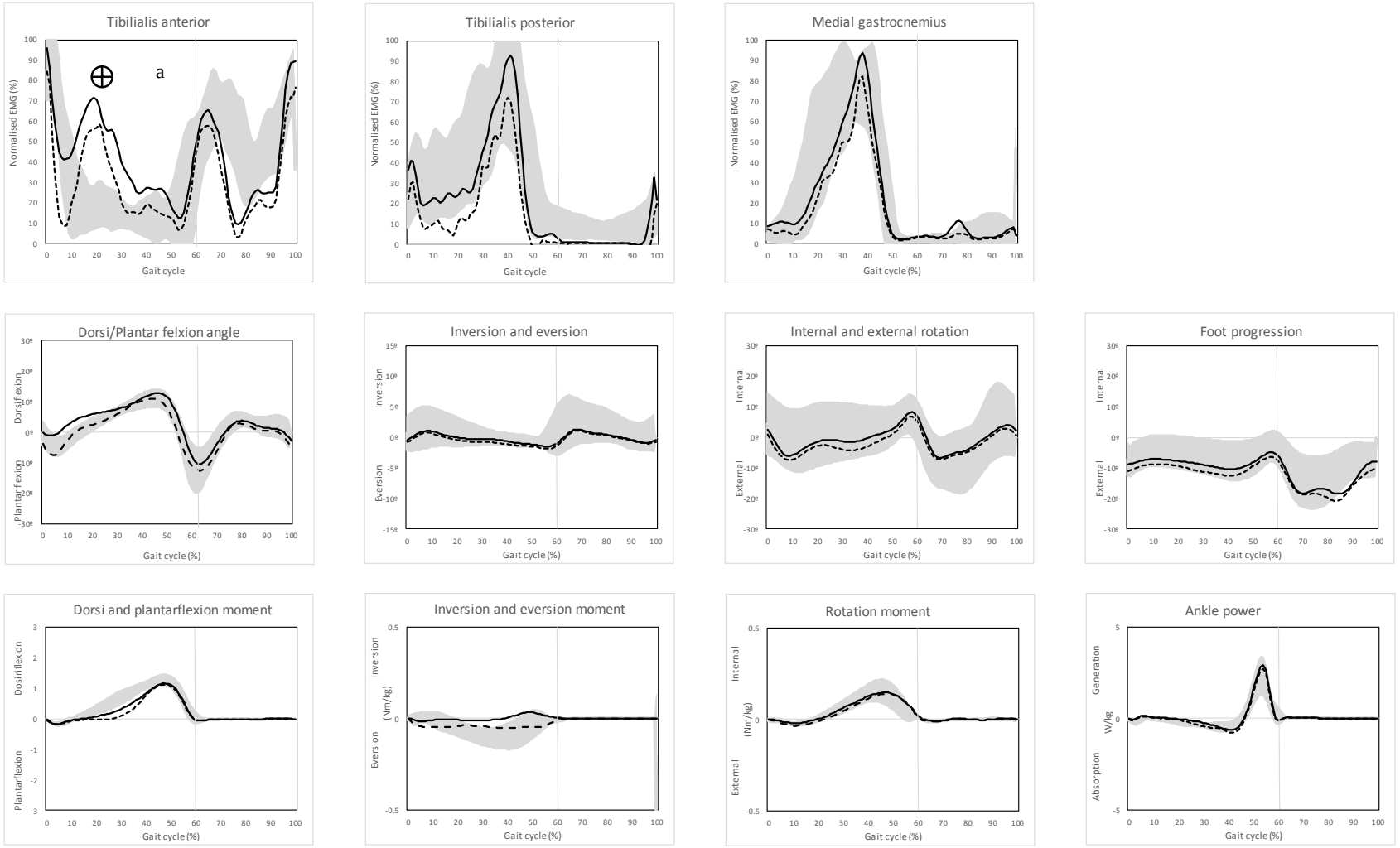


Figure 8.1 Healthy01 walking barefoot compared to the definitive normative database (right side).

All stroke patients in these case series showed spatio-temporal asymmetries evident in both the kinematic and kinetic gait deviations. A higher limb index (the ratio of the stance time of the ipsilateral foot and the contralateral foot), longer stance and longer gait cycle length of the sound side for stability were common across all 3 participants with stroke. Therefore, the timing of the peaks in all of the gait data (EMG, kinematics and kinetics) were slightly delayed on the sound side compared with normative data. A drop foot is clearly shown in two cases characterised by the plantarflexion through the swing phase.

8.4.1 Case 2: STK01

A 57-year-old man, with a mass of 96 kg, height of 163 cm, a Fugl Meyer lower limb score of 24 and a 10 year history of stroke presented with the most left spastic foot drop foot amongst the three participants. The self-selected walking speed was 0.5 m/s. Based on functional mobility assigned on the basis of walking speed (Schmid et al., 2007), he is classified as having limited community ambulation (speed between 0.4-0.6 m/s). His concerns involve tripping, ankle instability and quick exhaustion. He was prescribed with a hinged plastic AFO with plantar flexion stop and functional electrical stimulator (FES). He prefers FES due to the better cosmetics and easy donning but could not use FES for a long time due to fatigue.

His self-selected walking speed (barefoot) was slower than the slowest speed of the normative database (0.6 m/s). As a result, this patient's data is compared to the slowest normative data available. Compared to this normative reference, a different timing and magnitude of activation are evident in the three muscles on both the paretic (red, denoted by (a-i) in Figure 8.2) and sound sides (green, denoted by (j-m) in Figure 8.2). Particularly on the paretic side, the ankle is in the plantarflexion (point (b) in Figure 8.2) outside the normal bands through the gait cycle with slight inversion and internal rotation outside the normal

bands from stance to early swing- (point (g) in Figure 8.2) - equinovarus. There is an increase in the inversion moment (point (h) in Figure 8.2) and the low ankle power (point (i) in Figure 8.2) on the paretic side. Activity of the tibialis anterior is absent during early swing and the level of activity is lower than the normal SD band through the swing (see point (a) in Figure 8.2). The tibialis posterior and medial gastrocnemius are active through the gait cycle, including in the swing (point (d, f) in Figure 8.2) compared to inactivity in the normal reference. There are additional bursts of tibialis posterior activity between 0-5% of the gait cycle and a fluctuating level of activity continuing through the stance, which are not present in the normal reference (point (c) in Figure 8.2). This is suggestive of some clonus-like activity within the muscle. In the medial gastrocnemius, the timing of the peak activity occurs at 60% of the gait cycle, whereas the peak in the normative reference occurs at 40% (point (e) in Figure 8.2). Many of the graphs on the sound side appear abnormal in the second half of the gait cycle but most of this can be attributed to the sound side stance being much longer in this participant than in the normative database.

Gait was also measured when STK01 walked with a plastic hinged AFO with a plantarflexion stop (Figure 8.3). The plantarflexion range is absent (as expected) through the gait cycle (point (h) in Figure 8.3) whereas the ankle is in a plantarflexed position between 0-15% and 50-70% of the gait cycle in our normative data. At approximately 50% of the gait cycle, the ankle is in eversion outside the normative SD bands (point (j) in Figure 8.3) and in contrast to an inverted foot position when walking barefoot (Figure 8.2). The dorsiflexion moment is lower than the normative database both with an AFO (point (k) in Figure 8.3) and walking barefoot. The paretic ankle power is lower than the normative reference (point (m) in Figure 8.3) and point (i) in Figure 8.2. The maximal amplitude of EMG is reduced in three muscles for both sides (denoted by (a-d, n) in Figure 8.3). The

additional medial gastrocnemius activity in early stance disappears and continuous bursts of tibialis posterior activity are less pronounced, compared in Figure 8.2.

EMG analysis on the paretic side demonstrated that these kinematic and kinetic features are probably caused by continuous activity of the tibialis posterior, prolonged activity of the medial gastrocnemius and low activity of the tibialis anterior in swing as the plantarflexion is maximal at that point. The treatment plan for STK01 may aim to reduce the excessive activity of the tibialis posterior, such as botulinum toxin injection in the tibialis posterior, and assist dorsiflexion.

The kinematic data thus confirmed that the AFO has had an effect on the mechanics of walking (Figure 8.3). The paretic ankle is in a much better position (reducing the dorsiflexion moment and power which may manifest in EMG because muscular activation is not rquired to support the joint when the AFO takes over this function at this stage in the gait cycle) and there are also clear improvements in the sound side ankle kinematics. Despite this, the kinetic data show less evidence of changes. Perhaps most interestingly, the EMG activity particularly in the tibiailis posterior but also to a lesser extent in the gastrocnemius suggests some suppression of the clonus-like fluctuations in the signal, suggesting that the AFO has a neurological as well as mechanical effect.

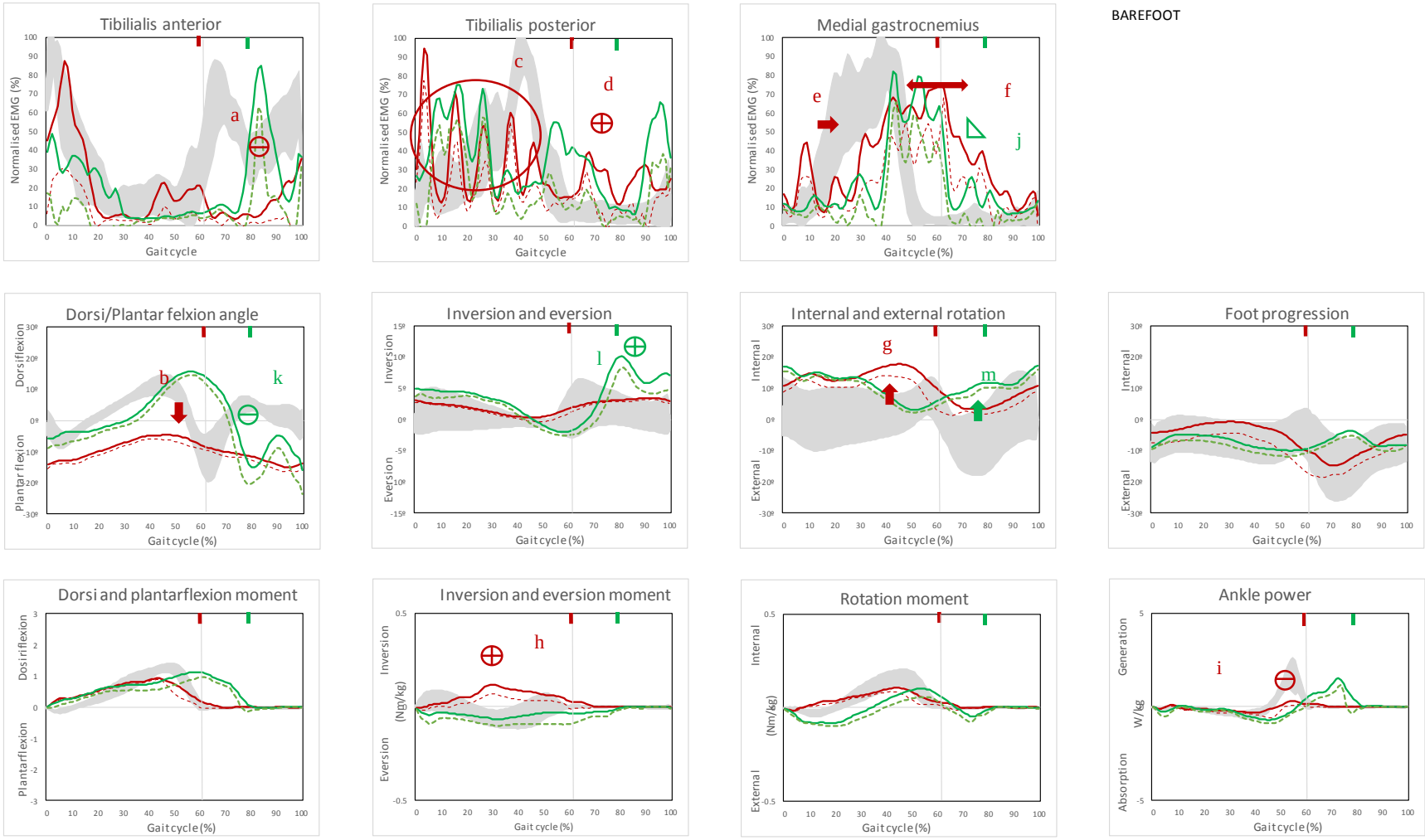


Figure 8.2 STK01 walking barefoot

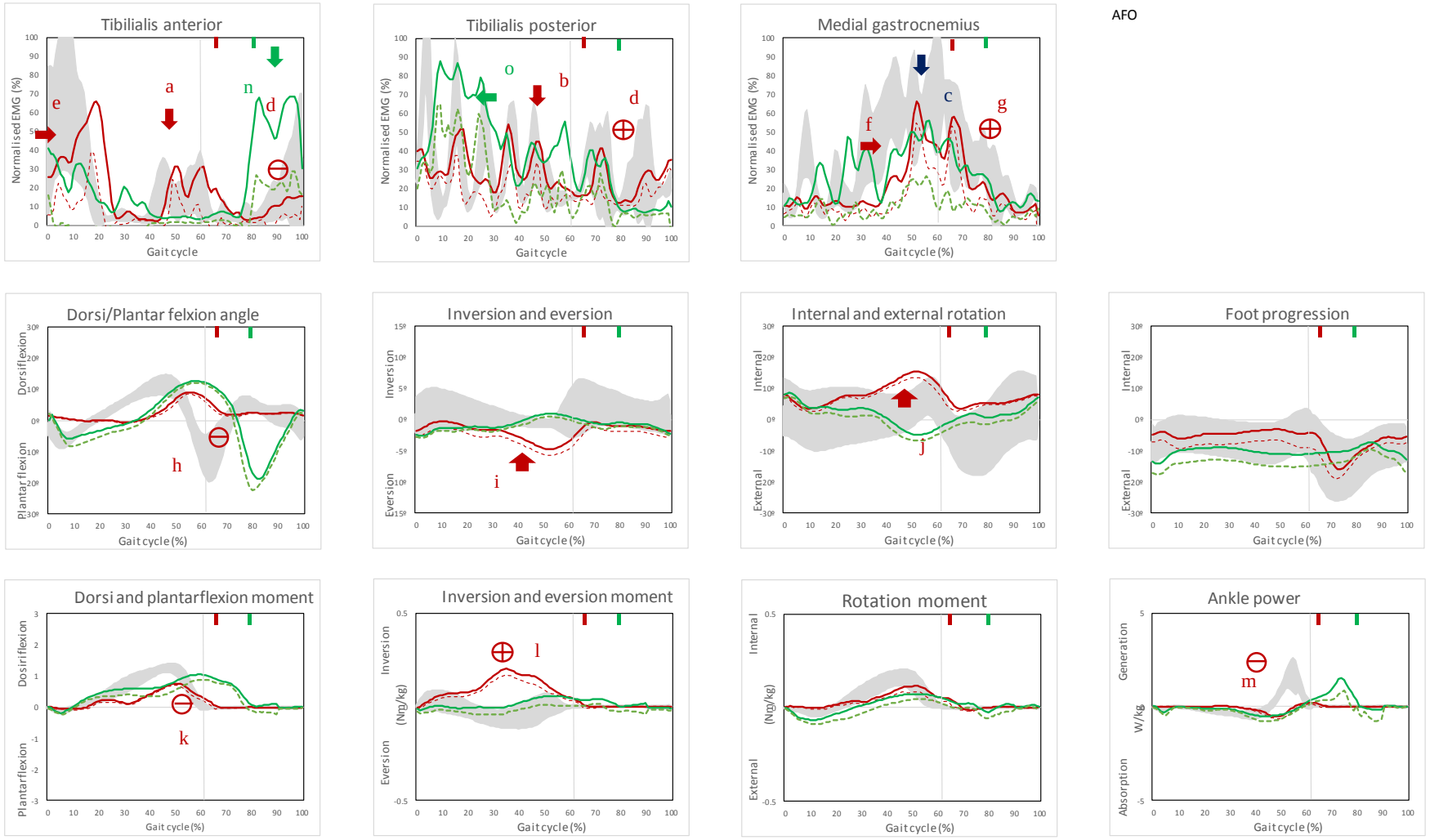


Figure 8.3 STK01 walking with AFO

8.4.3 Case 3: STK02

A 69-year-old man, with a mass of 86 kg, height of 170 cm, a Fugl Meyer lower limb score of 22 and a 43 year history of stroke presented with right foot drop foot. The self-selected walking speed (barefoot) was 0.8 m/s. Based on functional mobility assigned on the basis of walking speed (Schmid et al., 2007), he is classified as having limited community ambulation. His concerns involve tripping and ankle instability. He was prescribed a hinged conventional AFO with a plantar flexion stop. He usually uses it for long distant walking or a long standing task.

This patient's data are compared to the normative data at 0.94 m/s (25% reduced self-selected speed obtained from healthy participants). Compared to the normative reference, the different timing and magnitude of activation is evident in the three muscles on both the paretic (red, denoted by (a-i) in Figure 8.4) and sound sides (green, denoted by (g,h,k) in Figure 8.4). A small degree of plantarflexion outside the normative SD bands in the swing phase on the paretic ankle can be seen (point (f) in Figure 8.4) and at forefoot initial contact. The activity of the tibialis anterior in the swing phase and the transition from swing to stance is lower than the normative reference (point (b) in Figure 8.4) and its peak occurs in the stance before toe-off. This is in contrast to the normative data, where tibialis anterior activity occurs after the toe-off (point (a) Figure 8.4.). Moreover, the medial gastrocnemius is active through the swing phase and transition from the swing to stance phases (point (e) in Figure 8.4) and the tibialis posterior is active at 90% of the gait cycle (point (d) in Figure 8.4) outside the SD bands of the normative reference data. The ankles on both sides are more inverted than the normative SD band for between 30-50% of the gait cycle and remained in that position through the gait cycle (point (g) in Figure 8.4). On the paretic side, the tibialis posterior activity between 30-50% is lower than the normative range but the tibialis anterior is active (outside the SD band) at this period (point (a) in Figure 8.4).

STK2 has been prescribed with a conventional (leather straps, stainless steel rods and leather shoes) hinged AFO with a plantar flexion stop. He only uses it when walking for a long distance due to donning difficulties and cosmetic acceptance of the orthosis. The paretic ankle has no range of plantarflexion (point (a) in Figure 8.5), compared to the plantarflexion occurring during toe-off and early stance in the normative reference data. The activity of the tibialis anterior is lower than the normative reference band in swing (point (b) in Figure 8.5) but the activity of the medial gastrocnemius is higher than the normative database in swing (point (d) in Figure 8.5). These are similar to Figure 8.4. In the stance phase, the eversion moment has higher variability than the normal SD bands (point (h) in Figure 8.5) and walking barefoot (which is reflected to a lesser extent in the plantarflexor moment). A relatively higher peak and SD of the EMG is found in the activity of the tibialis posterior than in the normative reference bands (point (c) in Figure 8.5) and Figure 8.4, and there is considerably greater variability when walking with the AFO. On the sound side, the peak amplitude of all muscles increases compared to Figure 8.4.

The pathological kinematic and kinetic gait deviations of STK02 are mild/moderate. Small degrees of plantarflexion in swing and a small degree of inversion and internal rotation in part of the gait cycle were identified but a dysfunction in muscle activation was clearly seen on the paretic side (pronounced difference): low activity of the tibialis anterior and continuous activity of the medial gastrocnemius through swing and abnormal tibialis posterior activity prior to initial contact. The treatment plan is likely to be dorsiflexion assist to overcome medial gastrocnemius activity in swing.

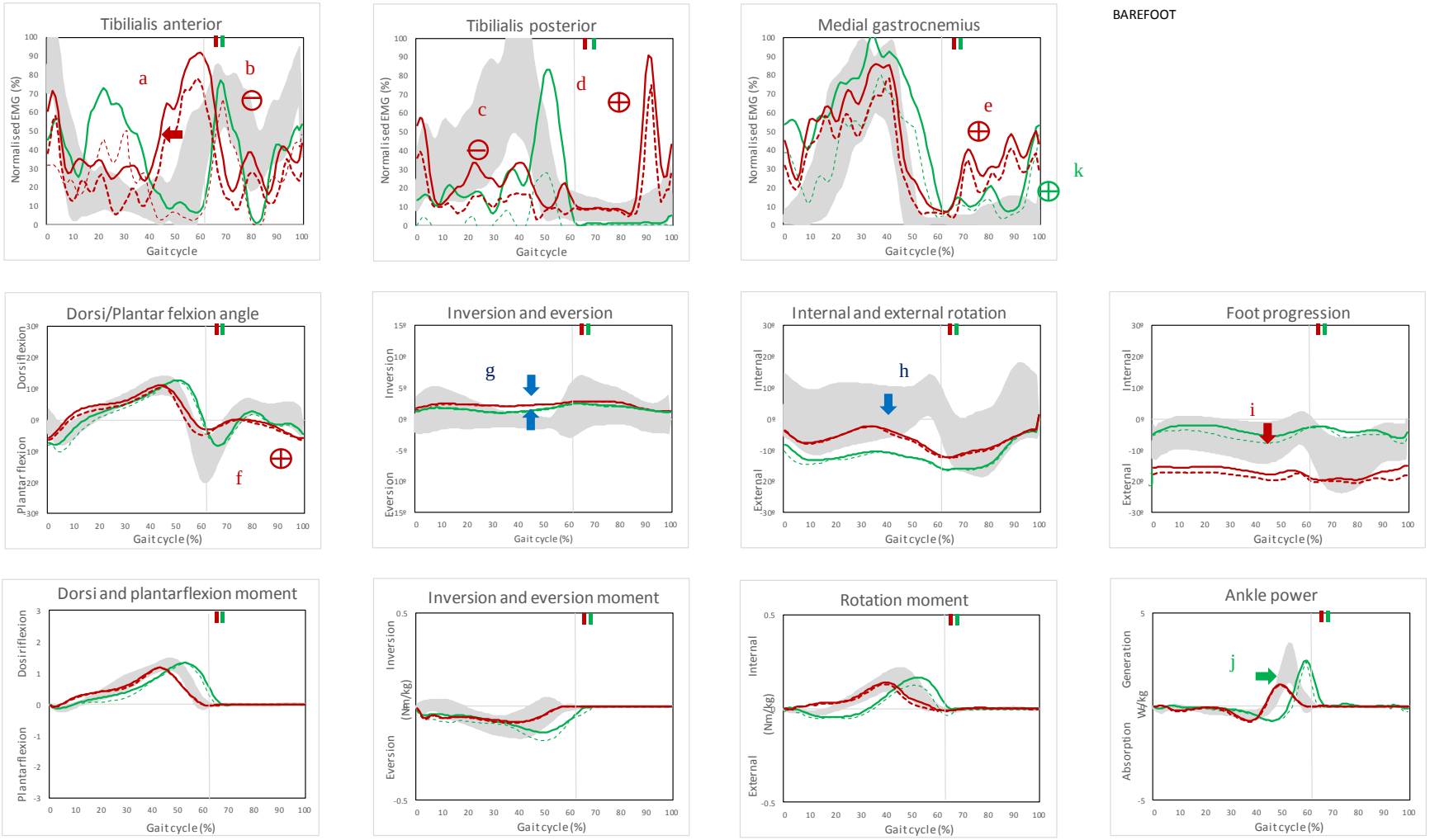


Figure 8.4 STK02 walking barefoot

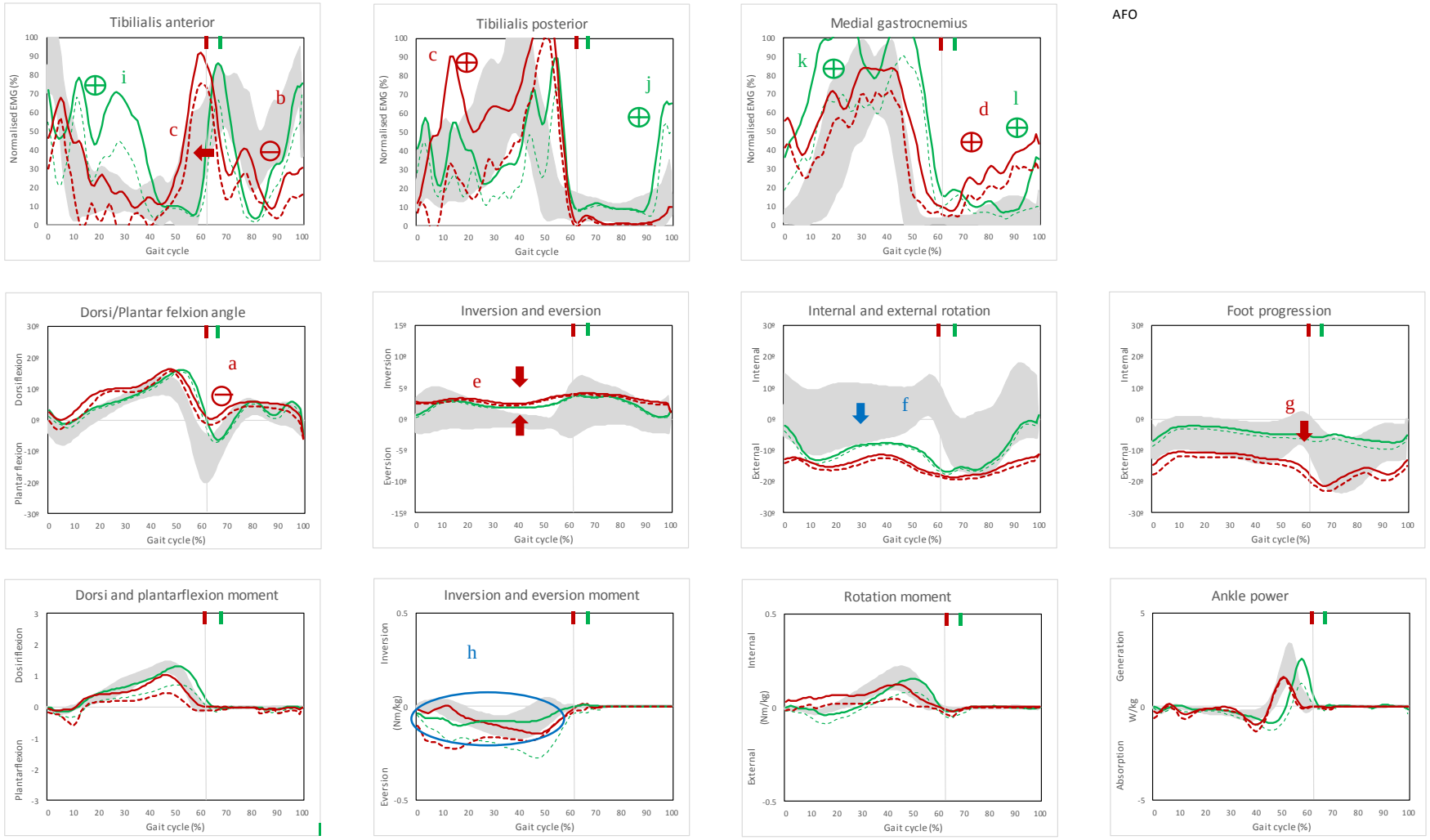


Figure 8.5 STK02 walking with AFO

In summary, the AFO improves the paretic ankle angle in swing in the sagittal plane and gait cycle length but appears to have little effect on the other kinematic parameters (Figure 8.5). Generally, the patterns of muscle activity for all muscles were changed by AFO to be more similar to normal activity, and the increase in ankle power indicates the AFO may be beneficial for this patient in prolonged walking/standing tasks. There is considerably greater variability in the coronal plane ankle moments (and to a lesser extent in the sagittal plane moments) with the AFO. This may be a reflection of some variability in the gait pattern when walking with an AFO which is rarely worn and with which the participant is thus unfamiliar. It does suggest that the AFO may be eliciting some inappropriate and variable activity in the tibialis posterior on the sound side which may be contributing to this. To reduce this variability in the tibialis posterior, a medial arch support integrate on the foot plate may be beneficial.

8.4.4 Case 4: STK03

A 64-year-old man, with a mass of 68 kg, a height of 168 cm, and a Fugl Meyer lower limb score of 20 and a 6 year history of stroke presented with right foot drop foot. The self-selected walking speed (barefoot) was 0.6 m/s. Based on functional mobility assigned on the basis of walking speed (Schmid et al., 2007), he is classified as having limited community ambulation. STK3 does not have active isolated dorsiflexion or a range of eversion on the paretic side when the subtalar joint is stabilised during physical assessment. His concerns involve ankle instability and discomfort of the foot and ankle. He was prescribed with posterior leaf spring AFO but rejected this due to discomfort. Then he was given a soft fabric AFO which is used daily. He regularly wears his orthosis as he feels more stable when he does.

This patient's data were compared to normative data taken at 0.64 ±0.15 m/s. The plantarflexion on the paretic side at 60% of the gait cycle is less than the normal range (point (e) in figure 8.6) with lower dorsiflexion moment and power outside the normative band (denoted by (h,j) in figure 8.6). In swing, the pattern of the tibialis anterior EMG is similar to the normal database with low activity in early swing (point (a) in figure 8.6). The activity of the tibialis posterior is lower than the normative reference data in the stance phase (point (b) in figure 8.6). The activity of the medial gastrocnemius reaches a peak at 15% of the gait cycle but, in the normative reference data, the peak is at 40% of the gait cycle (point (d) in figure 8.6). The paretic tibialis posterior is active in the swing phase outside the normative SD bands (point (c) in figure 8.6).

When walking with a fabric AFO and shoe, his walking speed, gait cycle length and gait cycle time increase from 0.6 m/s to 0.7 m/s. On the paretic side, the ankle is in dorsiflexion during toe-off (point (a) in Figure 8.7) and the dorsiflexion in swing is higher than the normative reference data (point (b) in Figure 8.7) and the range of ankle sagittal and frontal movements increases compared with walking barefoot (Figure 8.6). The inversion moment both with the fabric AFO and walking barefoot is higher than the normative reference during 20-40% of the gait cycle (point (m) in figure 8.7) (Figure 8.6). The peak EMG amplitude of the tibialis anterior and tibialis posterior increase with the fabric AFO compared with walking barefoot (Figure 8). The EMG pattern of the medial gastrocnemius in stance is within the normative range in stance but outside the normative range in swing (point (h) in Figure 8.7). The activity of the medial gastrocnemius is prolonged compared with walking barefoot. On the sound side, the EMG of the tibialis posterior is almost constant across the gait cycle (point (g) in Figure 8.7). This low signal to noise ratio suggests a poor sensor placement and should not be interpreted.

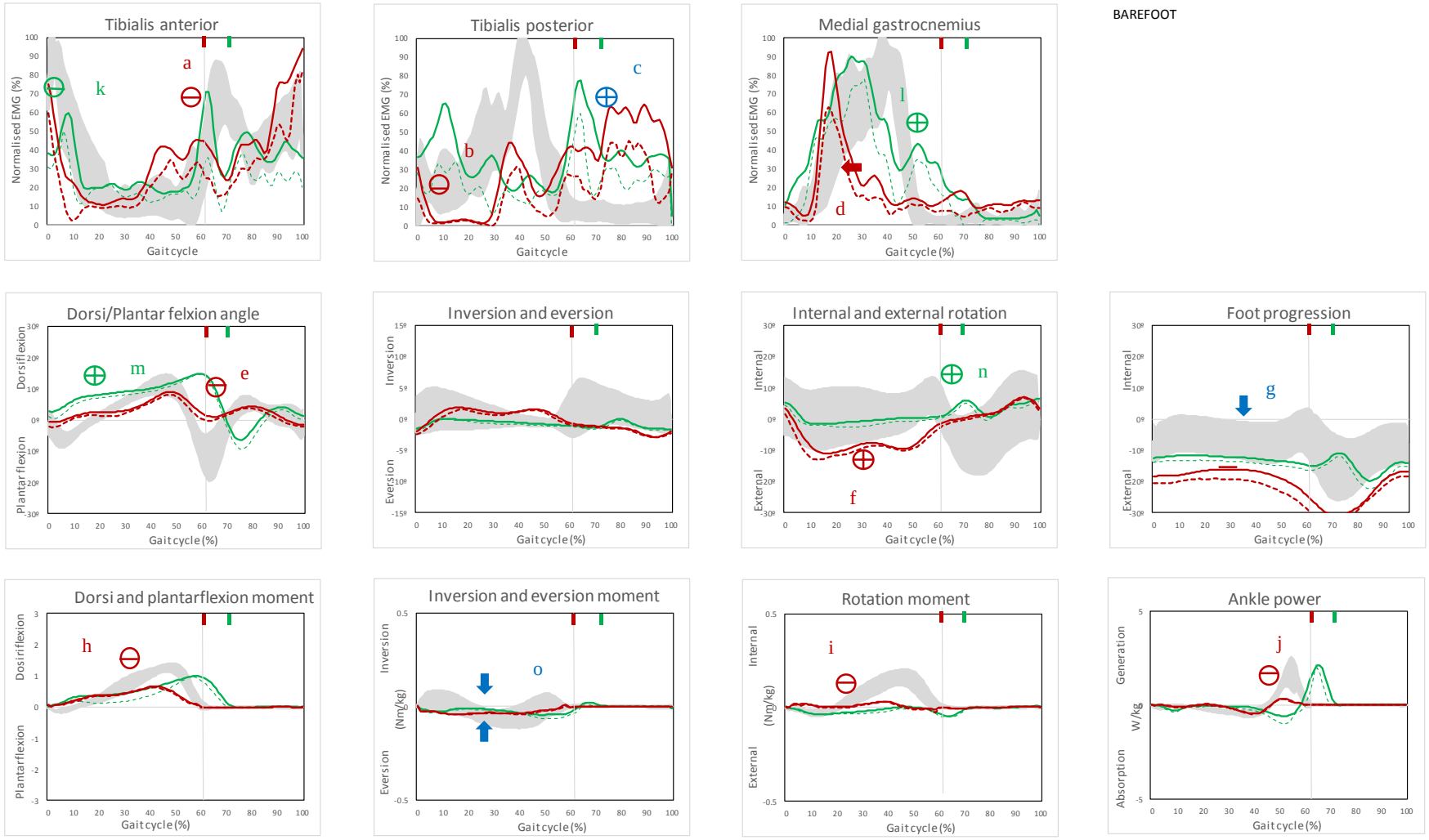


Figure 8.6 STK03 walking barefoot

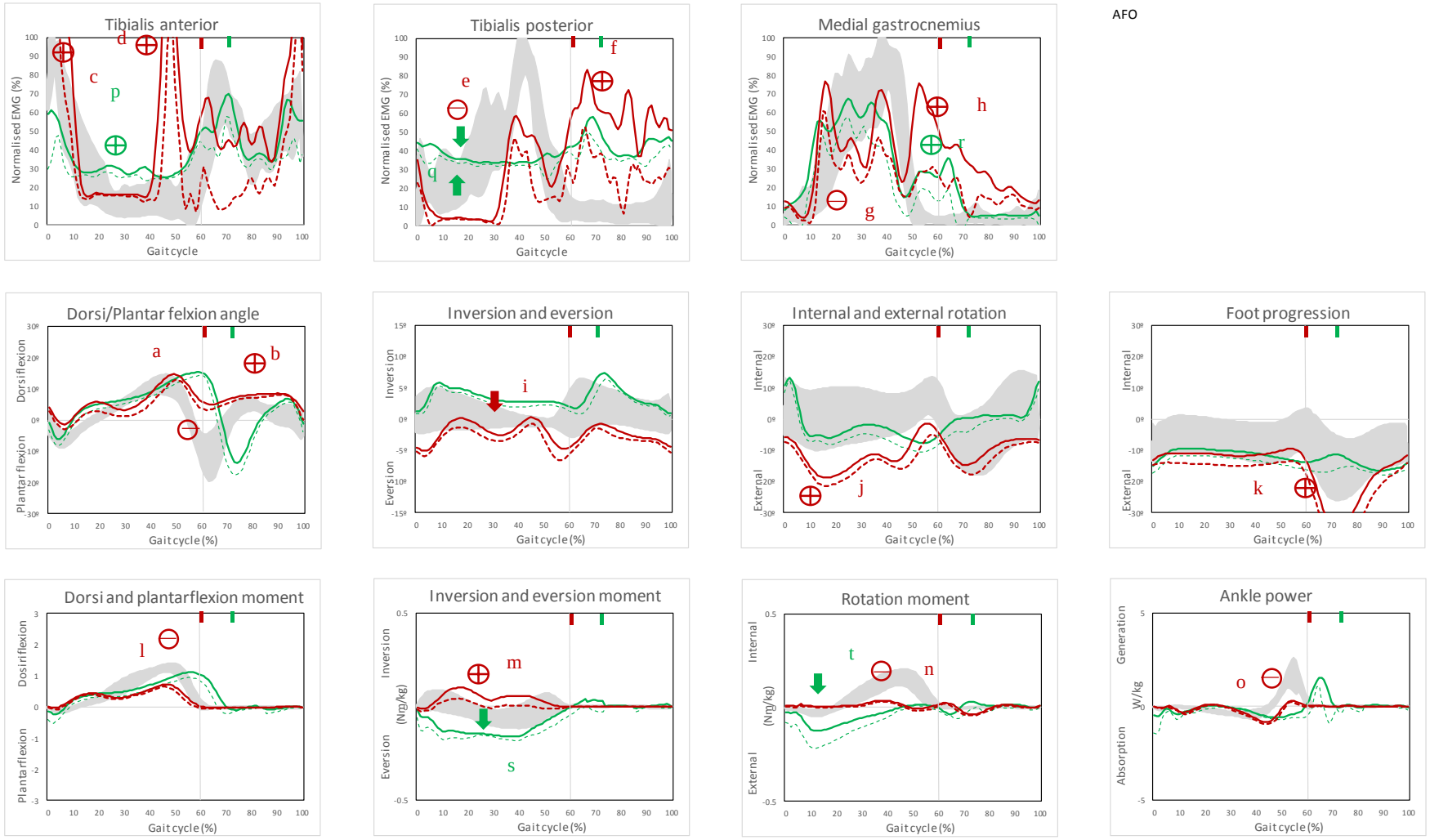


Figure 8.7 STK03 walking with AFO

Overall, this is a complex set of data. The AFO appears to have affected the coronal plane kinematics and kinetics of both the paretic and sound side, suggesting that it has made walking more abnormal (especially) in the coronal plane although there is evidence of improved dorsiflexion in swing on the paretic side. There seems to be evidence of more normal firing of the gastrocnemius through stance with the AFO but is accompanied by an increased magnitude of abnormal activity in the tibialis anterior and posterior. The signals from the tibialis posterior on the sound side suggest some measurement artefact. It may be that the orthotic prescription could be modified to produce similar improvements in clearance but without the coronal plane abnormalities that seem to be characteristic of the current prescription.

8.5 Discussion

8.5.1 Identification of EMG pathological changes

Using the protocols and definitive normative EMG profiles established in chapter 7, only one difference is found in the EMG profiles between an older healthy adult and the normative data. Only the activity of the tibialis anterior increases by 10-30% of gait cycle. The activity of the tibialis posterior and medial gastrocnemius is within the normative range of younger healthy adults at matched-speed. The obvious difference in tibialis anterior activity draws attention to more subtle changes in gastrocnemius activity and the sagittal plane ankle joint angles and moments, which might have been overlooked in the absence of EMG data. These features appear to represent a mechanism within the normal limits of walking, and a comparison with the data presented in Figure 7.12 (Chapter 7) suggests that this may be a reasonably common result of ageing. These similarities provide proof-of-concept for using our established protocol and corresponding normative data in that they give insights into the different mechanisms that make up the range of variability within healthy walking.

The overall differences in EMG between participants with stroke and normative reference data can be identified when comparing data from patients with normative reference data gathered using the proposed protocols at matched speeds. In the case series of patients with drop foot, the activity of the tibialis anterior is absent during early swing for all participants with stroke and low throughout the swing (STK01-Figure 8.2, STK02-Figure 8.4). The tibialis posterior and medial gastrocnemius are inappropriately active in swing: prolonged activity from stance to swing (tibialis posterior of STK03-Figure 8.6, medial gastrocnemius in STK01-Figure 8.2) or an additional burst of activity (both muscles in STK01-Figure 8.2 and STK02-Figure 8.3) and inappropriate timing during stance (tibialis posterior in STK01-Figure 8.2 and STK02-Figure 8.4, medial gastrocnemius in STK03-Figure 8.6). Differences between the EMG on the sound side and normative data can also be seen. Further, the differences in EMG seen between patients with stroke and normative reference data reflect both neuromuscular impairment (e.g. drop foot) and biomechanical restriction (e.g. AFOs). This is evidence supporting the concept: EMG profiles collected using previously established protocols can detect changes in these muscle activities due to stroke.

8.5.2 EMG reports and other gait data

a) Pathological changes of EMG contribute to gait deviation

In order for EMG usefully to support standard kinematic, kinetic and power CGA, the variability of the normative EMG reference data should be sufficiently low to allow changes in EMG to allow neuromuscular pathology to be detected. In patients with severe kinematic and kinetic impairment, differences in EMG data compared to normative reference data should be detectable, or the variability of the current normative data is too large compared to the effect size of the impairments and this would rule out the use of EMG within CGA.

The ability to detect pathological changes is the primary purpose of the normative database. EMG from a case series of participants with stroke clearly showed differences in muscle activity from the paretic and sound sides compared to the normative database, even in participants with mild impairment for whom kinematics and kinetics showed small differences (STK2-Figure 8.2). With the EMG reports, the clinician can verify whether the small changes in kinematics or kinetics are likely to be caused by neuromuscular impairment. Moreover, EMG reports can be used to assess the severity of the impairment. The stroke participants with the most severe motor impairment showed the greatest differences in EMG compared to the normative reference data than stroke participants with mild/moderate severity.

Gait deviations manifesting in kinematics and kinetics, such as drop foot, can be a result of either an absence or low level of activity of the tibialis anterior, or prolonged or abnormal activity: plantarflexors (either gastrocnemius or soleus), invertors (tibialis posterior, or sometimes the tibilalis anterior) or toe flexors (flexor halluscis longus/brevis and flexor digitorum longus/brevis) (Ward, 2014). EMG can assist in the identification of these causes and also other pathological muscular dysfunctions occurring in gait, such as clonus. Perry et al. (1978) reported the absence of tibialis posterior activity in 10 out of 20 patients with equinovarus, and in 7 of the patients with varus and 6 of the patients with equinus, the tibialis posterior may be active in late swing and active for shorter periods, resulting in muscle imbalance and foot-ankle deformity. Based on an individual's requirements for gait data to aid foot clearance and the preposition of the foot in swing and loading of the stance leg and ankle stability for stance, the EMG reports could specify target muscles for treatment such as physiotherapy (selective and intensive exercises or electrical stimulation), chemical denervation-botulinum toxin injection (spastic management) and prescription (e.g. stiffness properties) of ankle foot orthosis to compensate for weakness and provide stability. EMG

reports could also assist surgical plans as the EMG patterns only slightly changed after surgery, so muscle activation patterns after surgery can be predicted (Renzenbrink et al., 2012). Therefore, the EMG measurement of these muscles should be implemented in the CGA, as these data are important for the treatment plan but should not be used as the only method to assess gait performance.

b) Effect of ankle joint restrictions

This section examines whether or not EMG varies from normative data in the presence of a mechanical restriction to the kinematics of walking i.e. AFO which limits plantarflexion. Indeed, in participants with AFOs, EMG differences can be seen at points in the gait cycle when the AFO would be providing the greatest biomechanical/kinematic restriction.

The hinged AFOs in this case series effectively prevents drop foot and provides mediolateral stability (reduced inversion and internal rotation), although the effects on EMG profiles vary, depending on the individual. Previous studies have reported a decrease in the EMG amplitude of the tibialis anterior and concerns about muscle dystrophy after long term use (Hesse, Werner, Matthias, Stephen, & Berteanu, 1999; Miyazaki, Yamamoto, & Kubota, 1997; Yamamoto, Miyazaki, & Kubota, 1993). As our data are drawn from the case series, it is unlikely to draw conclusions for the whole hemiparesis stroke population. However, the case series emphasised that hinged AFO may not always reduce the EMG amplitude of the tibialis anterior or medial gastrocnemius, depending on an individual's condition: STK01 rarely uses the AFO (reduced EMG amplitude) because he has another device (FES) but STK02 uses AFO (increased EMG amplitude may be a result of stabilised foot and ankle) for long standing or walking.

Moreover, the soft fabric AFO showed the effects on kinematics and EMG amplitude, despite the small amount of controlling force provided. Any changes to EMG seen when

walking with this device (as opposed to walking barefoot) may reflect changes in the way in which the participant is actively controlling his walking pattern. The compression around the ankle provided by fabric AFO may support the foot-ankle structure and sensory information. The biomechanical explanation for this patient is still unclear, based on the available information.

EMG data can inform clinicians about how muscle activity changes after orthotic or other intervention. Different types of orthoses may have different effects on muscles activity. Further, because the reason for impairment between individuals with the same conditions (e.g. drop foot) may be different among the patient group, EMG could potentially provide useful information about patient specific impairments which may help clinicians select a treatment. However, recent literature provides very little evidence on the effects of orthoses on muscle activity from which to draw conclusions about suitability of particular interventions or type of orthosis for any particular group of patients. Therefore, the exact way (e.g. which measures of EMG are necessary and/or sufficient to demonstrate, appropriateness, efficacy and mechanisms of effect of any given treatment) in which EMG could be used to support clinical decision making is not yet established. The protocol derived from this PhD is therefore the first step in determining if differences in muscle activation between patients and healthy normative data can be seen and if these differences correspond to, and help further explain, kinematic and kinetic gait deviations seen. Further future studies will be needed to determine which measures of EMG best capture particular impairments and treatment effects in order to facilitate the further use of EMG in patient' assessment and may lead to establishment of EMG use for orthotic prescription and other intervention

8.5.3 Limitation

With the use of peak normalisation and available techniques to process EMG data, the actual changes in EMG amplitude, which may be the result of the structural rearrangement of the motor units occur after stroke (Dattola et al., 1993; Hara et al., 2004), are not shown in these case series. Our protocol can only compare the patterns and relative amplitude to the peak of the EMG signals during gait. Despite this limitation, the protocols can be employed to detect pathological changes and the effects of intervention.

The quality assurance in EMG is great importance. Despite the use of a muscle stimulator and voluntary contraction to ensure measurement of targeted signal, there is a possibility of having a low signal to noise ratio during gait. The normalised EMG could be misinterpreted as seen in the fine-wire EMG of the tibialis posterior in STK03. Therefore, the signal should be processed for quality checking while the patient is present to allow re-application of the electrodes to ensure good quality data; for example, the low signal to noise ratio may result in similarity of processed EMG between resting and walking and unclear active and inactive periods of muscle activity during walking. Skilled personnel are required to carry out collection, processing and analysis.

Not all differences in EMG are manifested in kinematic deviations as illustrated in the case study of STK03 and several limitations have been identified within this case series with regard to the utility of EMG. For example, it is unclear how the delayed burst of gastrocnemius (STK01-Figure 8.2-3) or the short burst of tibialis posterior activity in late swing (STK02-Figure 8.4) affects the gait pattern. This may be a result of the increased inversion and may be underestimated in the current model.

8.5.4 Future studies

This study is only a case-series to provide proof-of concept for the use of established EMG protocols. Therefore, cohort studies are required to produce a better understanding of gait data in patients with stroke or the evaluation of intervention in other pathologies. The larger sample may reveal more potential uses of EMG signals for different levels of impairment, such as identification for surgery and botulinum toxin injections. The larger sample may also raise other issues with regard to EMG collection and interpretation which have not been identified in these case series. Moreover, the EMG protocols could be evaluated in other pathological groups, such as children with cerebral palsy, to investigate the sensitivity of the protocols to other pathological changes. The range of speeds is limited, as one stroke participant walked slightly slower than our slowest database (50% reduced from the normal speed) so part of the difference may be related to speed. Future studies may include the collection of other ankle muscles playing important roles in gait, such as the peroneus longus, as this is one of the main evertors during motion.

8.6 Conclusion

The previously established EMG protocols and reports can detect pathological changes in muscle activity in patients with stroke and provide additional information to kinematics and kinetics for the assessment of severity and evaluation of the intervention. This insight information could allow an optimised targeted treatment plan and design of the orthosis in the future when more information has been gathered. Therefore, EMG reports of the tibialis posterior, tibialis anterior and medial gastrocnemius are recommended in CGA for patients with neuromuscular disease.

Chapter 9 Synthesis, clinical implications and conclusions

The aim of the work presented in this thesis is to develop collection, analysis and presentation protocols for incorporation of both fine-wire and surface EMG measurements into clinical gait analysis (CGA) for patients with neuromuscular impairments. This final chapter combines the results from the series of experiments conducted to synthesise the major findings that have been obtained.

9.1 Synthesis of findings

It has been proposed that EMG can assist kinematic and kinetic measures during gait by facilitating assessment of severity, monitoring progress and prediction of the outcome of intervention (Baker, 2006). In order to achieve this however we must: (1) establish standardised collections, processing and presentation techniques, (2) collate normative reference data, (3) understand how EMG profiles relates to normative kinematic and kinetic data, (4) understand how differences from normative data can be identified.

Although many studies have reported EMG of various muscles during typical walking, up until this PhD work, the EMG reports were in a variety of formats including raw representative data, on-off patterns, linear envelope and frequency analysis as examples. For clinical gait analysis, presentation of EMG in the same format as kinematics and kinetics (with time varying amplitude) will aid clinical interpretation. There has also not been a review of this literature which attempts to understand the level of consensus amongst individual studies on EMG profiles (with time varying amplitude) in healthy populations. Therefore the systematic review (Chapter 3) aimed to synthesise normative EMG profiles from a number of studies to determine if the consensus exist based on variability for reference in CGA.

The systematic review showed a wide range of variability between studies in lower limb EMG profiles, a lack of studies in deep muscles (such as tibialis posterior which play important roles in foot posture and gait), no standard recommendation for use of fine-wire EMG in terms of acquisition and processing (compared to the surface EMG) and a range of methods for EMG normalisation. This variety of collection and analysis techniques resulted in large variability ranging from 16% to 49% of peak value, in the current literature base, of EMG profiles for the same muscle between different studies. EMG profiles of superficial muscles (gastrocnemius, soleus, vasti and hamstrings) were generally more repeatable than the other muscles (adductor magnus and peroneus longus). Therefore, a series of experiments were conducted to address these limitations with a view to providing a set of normative data for under-studied muscles (in this work, tibialis posterior) to facilitate the use of EMG for tibialis posteriors and with potential to extend the methodology to other deep muscles within standard CGA.

Given that being able to distinguish between normal and abnormal is one of the key requirements of any biomechanical measure taken for the purposes of clinical gait analyses (R. Baker, 2006), so the potential for normalisation schemes to reduce variability is of critical importance. Normalisation offers the potential for a better comparison of EMG between subjects and between sessions when there is a re-application of the electrodes. It is important to know how fine-wire and surface EMG are affected by different normalisation techniques for comparisons and interpretation.

Study 1 (Chapter 5) showed for the first time that mean and peak are the most appropriate techniques to normalise the fine-wire EMG in tibialis posterior and the fine-wire and surface EMG of the same muscles: tibialis anterior and medial gastrocnemius for CGA. Mean normalisation appears to be the most effective method to reduce variability and this is true across muscles, sensors and different measures of variability (standard deviation [SD],

standard error of measurement [SEM], coefficient of variance, variance ratio and coefficient of multiple correlation). Peak normalisation is equally good as there is only a small difference between them (less than 5% difference in quantitative result) and provides the same scale y-axis of the reported EMG profiles for all muscles. This aids a direct comparison between muscles, as the magnitude of active bursts across the gait cycle are in relation to the maximal level required during walking. Therefore, to form a normative EMG template for clinical gait analysis, peak normalisation is recommended.

The maximal voluntary isometric contraction technique has the potential to allow more meaningful interpretation of normalised EMG data in CGA. However, despite controlled positions of the lower limbs on a dynamometer, the technique was found to result in increased variability between subjects and sessions compared to non-normalised EMG, and this approach is therefore not proposed.

In addition to normalisation, variability of EMG profiles can be reduced using standardised collection, processing/analysis and reporting procedures. Recently available SENIAM guidelines (Hermens & Merletti, 1999) apply to surface EMG collection only and so similar guidelines are needed for fine-wire if EMG from deep muscles is to be included in CGA. Chapter 6 aimed to address several technical challenges of fine-wire EMG of tibialis posterior along with surface EMG, kinematic and kinetic data. There was uncertainty about whether fine-wire EMG was representative of the entire muscle (and therefore whether fine-wire EMG could validly be used with in CGA). There was also uncertainty whether the variability of fine wire signals was comparable to that from surface sensors and thus whether the same number of gait cycles were necessary to capture reliable data for an individual, variability/repeatability of the EMG signals between subjects and between sessions for the same subject. The results showed fine-wire EMG signals are not sensitive to where the sensor is placed in the muscle belly and that between trial variability of the fine-wire signals

was similar to that of surface signals of the same muscles with a high correlation coefficient. It was concluded that six gait cycles from either sensor type are needed to produce a mean that is representative for the individual. Minimal differences in signals recorded from either sensor type across a range of walking speeds in either the shape of the waveforms or their variability were seen.

Results from this series of experiments suggest the following data collection and processing procedures and recent recommendations for surface EMG. These are more specific than the guidelines for surface EMG (SENIAM) and are necessary to obtain EMG for deep muscles using fine-wire electrodes (synchronously with kinetic and kinematic gait data) with comparable sensitivity, repeatability and variability to that obtained from surface EMG:

i) Fine-wire sensors should be placed using ultrasound guidance and confirmed with muscle stimulation;

ii) Data should be sampled at 3,000 Hz;

iii) Data should be processed with 2nd order Butterworth band pass filter (50-1,500 Hz), full wave rectified and an envelope created with a 9 Hz low pass filter;

iv) A minimum of six gait cycles are required for an ensemble grand average to be representative;

v) Data should be normalised to the peak signal across the gait cycle for all gait cycles;

vi) Graphs should be presented as time varying amplitudes (percentage of gait cycle) with speed matched normative data plotted as +/- one standard deviations bands for reference as is current best practice for kinematics and kinetics for clinical gait analysis.

i-iii are the recent recommendations from current literature (e.g. SENIAM, ISEK) and iv-vi are derived from series of experiments in this PhD thesis. Using these EMG collection and analysis protocols a set of normative data from healthy young adults walking at different speeds (ranging from 0.64-1.86 m/s) was collected for the tibialis posterior (fine-wire sensor). The systematic review identified only one other source of normative data for this muscle despite its potential importance in some common impairments seen in CGA services. EMG data from the tibialis posterior alongside surface EMG (tibialis anterior and medial gastrocnemius), kinematic and kinetic data was used to understand the function of these muscles during walking in healthy adults. When speed increases, EMG amplitude of lower leg muscles and range of kinetics and kinematics increase and timing of peak activity occurred earlier in gait cycle.

The change in muscle activity appears proportional to large changes in ankle power with speed whereas only small changes with walking speed are observed in kinetics and kinematics. This suggests that normative reference comparisons used in CGA needs to be made at speeds comparable to the patients' self-selected walking speed in order to be sure to detect differences in patients that are due to specific pathology not solely to differences in walking speed. This is thus the first study to suggest that speed matched normative EMG data is more important for clinical gait analysis than speed matched kinematic or kinetic data. This study thus provides the first normative dataset, for EMG of tibialis posterior and three dimensional kinematics and kinetics with variability between healthy subjects (SD) across different speeds.

Further sources of variability which may mask the ability to detect differences due to pathology may come from age of participants and speed of walking. Current CGA standard practice is to compare patients to normative data of healthy adults of any ages walking at self-selected speeds. Our studies showed that, unlike differences in EMG due to different

speeds of walking, differences between younger and older participants were smaller in tibialis posterior (shown here for the first time) and medial gastrocnemius. Signals from tibialis anterior by contrast did show some changes with age. Excessive activity during early stance and early timing of peak during swing phase were associated with an increased inversion moment through stance phase in older participants. This suggests that age dependence of activity throughout the gait cycle appears to differ across different muscles. Age matched data would appear to be more important for tibialis anterior than for gastrocnemius or tibialis posterior.

Finally a case series of EMG collections with participants with stroke was used to explore the proof-of-concept of how standardised EMG methods could be implemented in clinical gait analysis and the potential benefits of using EMG to support identification of reasons for gait deviations in CGA. A normative database collected using these established methods was effective to identify pathological features and changes of muscle activity in three participants with stroke when using ankle-foot orthosis (AFO). All studies illustrate how the inclusion of EMG data in a format comparable to kinematic and kinetic data can generate important clinical insights providing richer information for clinical decision making.

9.2 Clinical implications

According to the results from the systematic review (Chapter 3), there were only a relatively small number (9 out of 24 papers) of studies that used fine-wire sensors on superficial and deep leg muscles. EMG profiles of tibialis posterior (Murley, Buldt, et al., 2009; Murley et al., 2014) and popliteus (Davis et al., 1995), which are deep muscles, have been reported by only one research group. Analysis comparing fine-wire to surface EMG in tibialis anterior and medial gastrocnemius revealed fine-wire EMG is equivalent to surface EMG indicating fine-wire EMG can be interpreted similarly to surface EMG in clinical investigations. Further, our application of fine-wire sensors in three stroke patients and four older adults

provides proof-of-concept that this method of measuring activity of deep muscles is feasible within CGA.

As stated previously, EMG can assist kinematic and kinetic measures during gait to facilitate assessment of severity, monitoring progress and prediction of the outcome of intervention (Baker, 2006). Our investigations of EMG in a case-series of patients with stroke showed examples of how EMG can reflect the severity and nature of impairments. The stroke participant with the most severe motor impairment showed the greatest differences in EMG compared to normative reference data than that of stroke participants with mild/moderate severity. This patient showed pathological features in swing: absence of tibialis anterior, continuous activity of medial gastrocnemius and tibialis posterior and, in stance phase, a clonus like activity found in the tibialis posterior profile. Also, in mild/moderate impairments which were not apparent in the kinematic or kinetic data, the EMG highlighted pronounced impairments including continuous activity of the tibialis posterior and medial gastrocnemius in swing. This is one of the reasons why the patient requires an ankle-foot orthosis (AFO).

Our case-series further exemplified how EMG could aid in monitoring progress. In a patient who has used AFO regularly for 40 years, EMG patterns of tibialis anterior across the gait cycle and medial gastrocnemius in stance phase (over activity in swing phase when walking barefoot) matched reference data well despite expectations that volitional activation in these muscles might have been diminished after long term use of AFO. While this, effectively, cross-sectional view of a patient after long term use of an AFO does not provide a direct time line of response to a course of treatment, the high correlation coefficients between sessions seen in Chapter 6 give confidence that repeated measures of EMG over time could be used to monitor progress in this way.

The reports of EMG and other gait data from all stroke patients in a case-series clearly showed the difference between walking with and without AFO, showing the outcome of intervention with illustrations that the EMG may became more or less comparable to normative reference data following orthotic intervention. Further, EMG provided insight into the muscular impairments underlying drop foot (either over excitation of plantarflexors or under activity of dorsiflexors). This information can be used to predict how a patient might respond to typical treatments such as use of AFO to prevent plantarflexion in compensation for absence of tibialis anterior activity or injection of botulinum toxin to reduce spasticity in one of plantarflexors. However, using EMG, particularly of tibialis posterior or other muscles associated with stabilising foot posture, to assess severity of impairment or to predict treatment outcomes might be further improved by use of a more comprehensive foot model to understand relative movements of forefoot, midfoot and hindfoot.

Currently, the EMG report provides the information of muscle activity at particular point of the gait cycle which can relate to kinematic and kinetic changes. These can be useful for targeted treatment plan for an individual but it does not provide information on how to select the appropriate orthotic design. This additional information on how the muscles changes in response to the orthotic or other intervention for an individual can be helpful in clinical setting for follow up and for further research. The protocol derived from this PhD may facilitate the further use of EMG in patient' assessment/follow-up and may lead to establishment of EMG use for orthotic prescription and other intervention.

9.3 Limitations

The process of fine-wire application is invasive and time consuming and is contraindicated for patients taking anti-biotic medication, anti-coagulant medication, and anti–platelet therapy; any immune deficiency conditions as the fine-wire insertion can cause the

destruction of several muscle fibres and may lead to further complications. As a result use of fine-wire EMG is not appropriate for all patients.

Using our protocol for monitoring progress and prediction of the outcome of intervention, the maximal amplitude of the normalised EMG profiles collected from the same muscle between sessions which requires re-application of electrodes cannot be compared as the signals are reported as proportion of the peak or mean value. Only the pattern of the profiles and the relative amplitude of different bursts of activity can be analysed.

9.4 Future studies

The protocols are designed specifically for distal ankle muscles: tibialis posterior, tibialis anterior and medial gastrocnemius. The reported sensitivity and repeatability of the fine-wire EMG and surface EMG may be muscle specific as the EMG signals can be influenced by quantity of fat and connective tissues or arrangement of motor units within the muscle. These properties between fine-wire and surface of EMG in proximal muscles may be different from our experiments because they may be covered by a larger amount of fatty tissue affecting the surface EMG. Therefore similar experiments should be carried out on other deep muscles, such as peroneous longus and adductor magnus, before incorporating in CGA.

The interpretation of EMG signals is often driven by clinical questions relating to the impairments which may be affecting walking. Therefore, most clinical users focus on the function of individual muscles separately; in contrast research applications which use multichannel EMG aim to develop an understanding of coordination and synergies (Frigo & Crenna, 2009). The studies in this PhD have shown it is feasible to include fine-wire EMG of tibialis posterior in CGA with stroke patients and those differences in tibialis posterior activity can be seen when compared to a normative reference dataset. A series of studies (Crenna, 1999; Crenna, 2003; Crenna & Inverno, 1994; Frigo & Crenna, 2009) have

demonstrated different approaches to use of EMG to help identify a range of impairments including paresis, spasticity and co-contraction. The discussion of how to identify and clinically analyse these impairments were beyond the scope of this review but future studies could look to apply these approaches to the analysis of tibialis posterior data or use of fine-wire application in other deep muscles in order to understand/identify impairments.

9.5 Final conclusion

To answer the overarching purpose of this work, our studies show surface and fine-wire EMG are comparable and can be used in CGA to identify even mild impairments in neuromuscular activation which are not apparent in kinematics and kinetics profiles. The established protocol in this study and corresponding dataset are regarded as a guide for instrumented gait laboratories to collect their own normative data and clinical interpretation for CGA in patients with neuromuscular disorders. A case series of patients with stroke provided proof-of-concept of the utility of EMG is useful in providing additional information about the muscular contributions to observe gait biomechanics. This could assist treatment plan by assessment of severity, monitoring progress and prediction of the outcome of intervention.

Appendix 1 Search strategies for systematic review

Cochrane Pubmed(medline) (from , Academic search premier(ENSCO): CINAHL, Medline, Sportdiscus, and Library information science and technology abstract, Web of knowledge.

1. electromyogra* or EMG*or * EMG
2. (muscular adj./near/N2 activit*) or (muscle* adj./near/N2 acticvit*) or (muscular adj./near/n2 activation*) or (motor unit)
3. gait* or walk*
4. normal or healthy or able bod*
5. 1 or 2
6. 5 and 3 and 4
5. Limiter: human and English

2.AMED(Ovid)

1. [electromyography] or EMG*
2. (muscular adj./near/N2 activit*) or (muscle* adj./near/N2 acticvit*) or (muscular adj./near/n2 activation*) or (motor unit)
3. 1 or 2
4. [normal] or normal. tw or healthy or able-bod*
5. [walk] or [gait] or walk* or gait* or locomoti*
6. 3 and 4 and 5
7. Limiter: human and english

Mesh terms

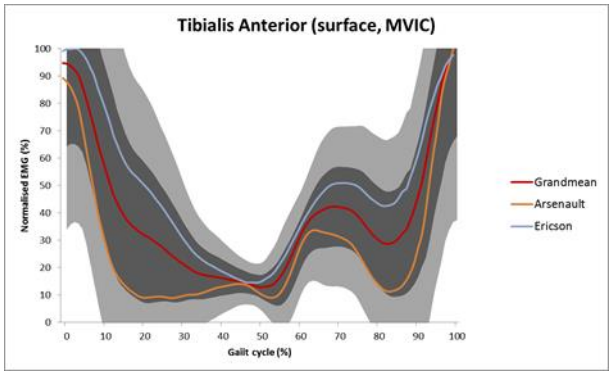
[Gait], [Electromyography],[walking][Human]

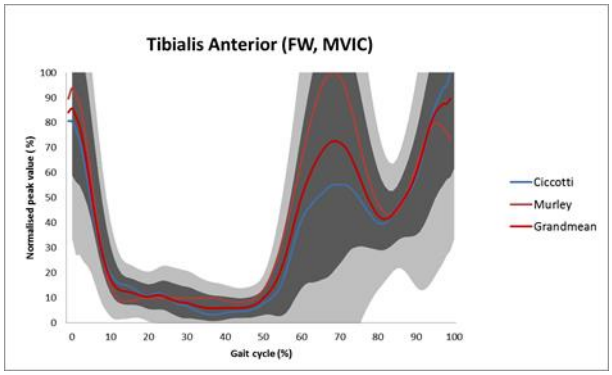
Appendix 2 Example of pooling graphs

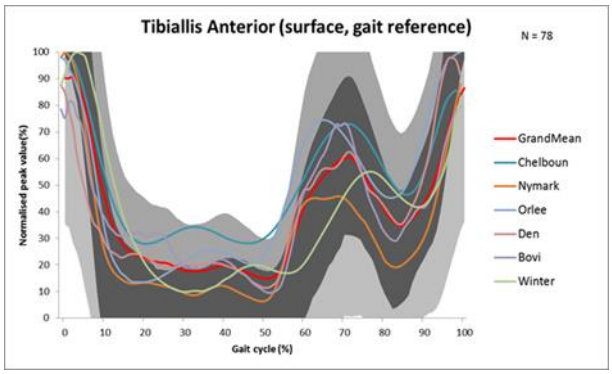
Tibialis anterior

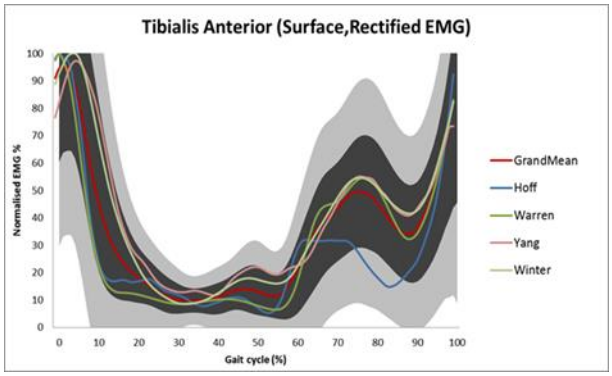
Author	Year	N	Variation	Condition	Speeds	Processing
Surface EMG, MVIC						
Arsenault	1986	8	SD	Floor	106.1 steps/min	Linear Env
Ericson	1986	10	Not mentioned	Floor	1.40 m/s	Linear Env
Total		**18**				
Fine-wire EMG, MVIC						
Ciccotti	1994	22	Not mentioned	Floor	1.50 m/s	Integrated
Murley	2009	14	SD	Floor	1.09 m/s	RMS
Total		**36**				
Surface EMG, gait reference						
Chelboun	2007	9	SD	Treadmill	1.30 m/s	Average (Peak)
Nymark	2005	18	SD at peak	Both	1.45,1.44 m/s	Linear Env (Mean)
Orlee	1995	10	SD	Floor	ssw	Linear Env (Mean)
Den otter	2004	9	Not mentioned	Treadmill	0.83	Linear Env (Peak)
Winter	1987	12	SD	Floor	ssw	Linear Env (Mean)
Bovi		20	SD	Floor	1.22	RMS (Peak)
Total		**78**				
Surface, rectified						
Hoff	2002	9	Not mentioned	Floor		Rectified
Yang	1985	11	Not mentioned	Floor	115 steps/min	Linear
Warren	2004	19	SD	Treadmill	1.12 m/s	RMS
Winter	1985	12	SD	Floor	ssw	Rectified
Total		**51**				

Note: MVIC =maximal voluntary isometric contraction normalisation
Gait reference = normalisation using the value obtained from walking such as mean or peak values
SD = standard deviation
Linear env = Linear envelope
RMS = root mean square
Ssw =self–selected walking speed





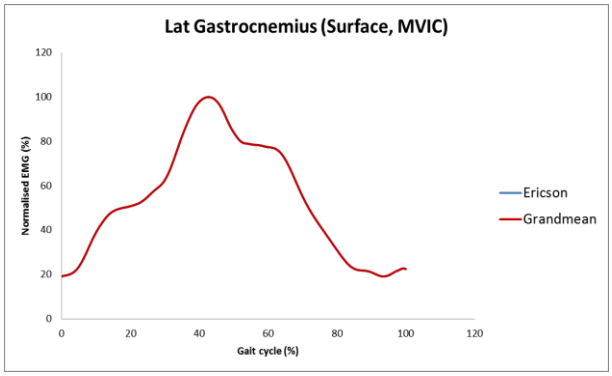


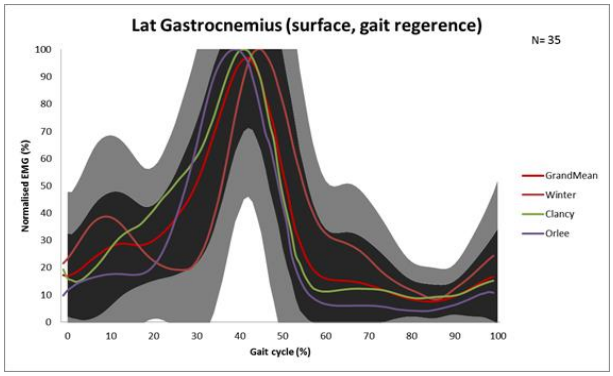


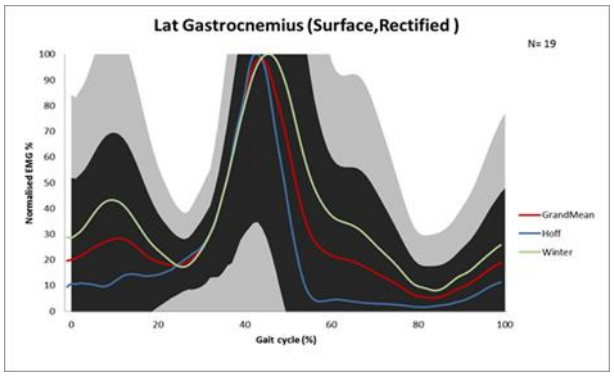
Lateral gastrocnemius

Author	Year	N	Variation	Condition	speeds	Processing
Surface, MVIC						
Ericson	1986	10	SD 16% at peak	Floor	1.40 m/s	Linear Env
Surface, gait reference						
Orlee	1995	10	SD	Floor	ssw	Linear Env (Mean)
Clancy	2004	15	Not mentioned	Treadmill	1.30 m/s	Average (Peak)
Winter	1987	10	SD	Floor	ssw	Linear Env (Mean)
Total		**45**				
Surface, rectified						
Hoff	2002	9	Not mentioned	Floor	1.25 m/s	Rectified
Winter	1985	10	SD	Floor	ssw	Rectified
Total		**19**				

Note: MVIC =maximal voluntary isometric contraction normalisation
Gait reference = normalisation using the value obtained from walking such as mean or peak values
SD = standard deviation
Linear env = Linear envelope
RMS = root mean square
Ssw =self–selected walking speed

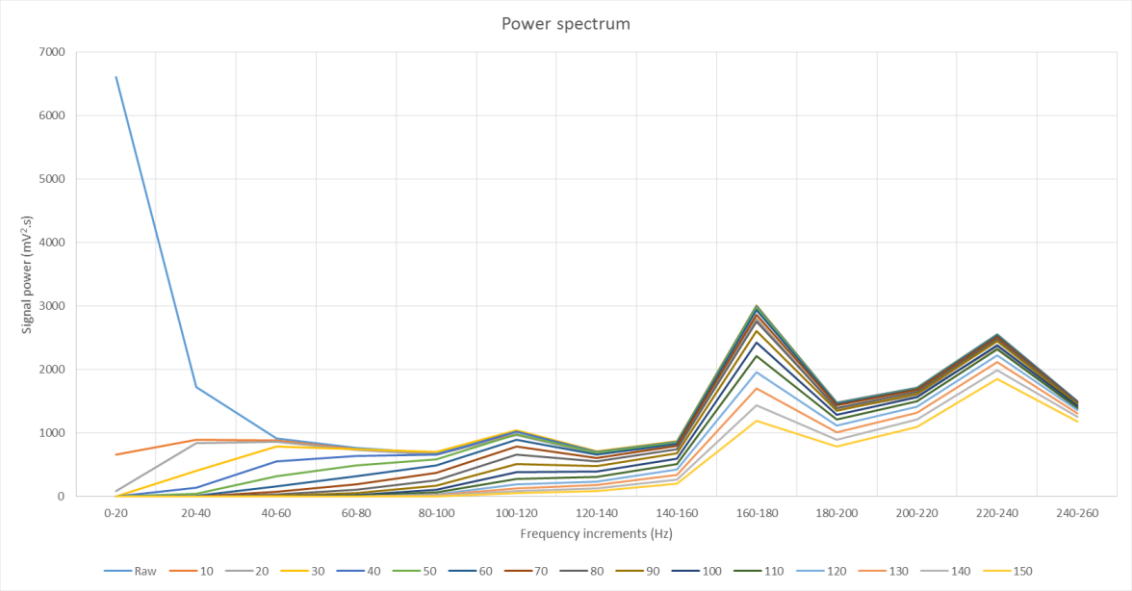

Lat Gastrocnemius (Surface, MVIC)
Normalised EMG (%)
Gait cycle (%)
Ericson
Grandmean


Lat Gastrocnemius (surface, gait regerence)
N= 35
Normalised EMG (%)
Gait cycle (%)
GrandMean
Winter
Clancy
Orlee


Lat Gastrocnemius (Surface,Rectified)
N= 19
Normalised EMG %
Gait cycle (%)
GrandMean
Hoff
Winter

Appendix 3 Power spectrum of the tibialis posterior

Power spectrum of a tibialis posterior EMG signal detected by a fine-wire sensor during standing. Each trace represents an EMG signal filtered by corresponding frequency values ranging from 10–150 Hz



The raw trace has more power in the range from 0 to 60 Hz. Then the shape of the power spectrum is similar to the filtered profiles. The spectrum line for 50 Hz of filtered EMG power appears to be a median, among other lines. Also, its peak at 160–180 Hz is similar to the raw EMG spectrum. This suggests that the main frequency components of EMG signals are retained after being filtered.

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